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# DEVELOPMENT OF AN EXOSKELETON MODEL IN A NEUROREHABILITTION PERSPECTIVE

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*"We may feel ill prepared to face the feared changes ahead, yet each of us can look back at our own lives and see countless times that something felt scary, hard and impossible. We were sure we wouldn't make it, and then we did. This is resilience - the willingness to persist, to learn from the experience, and to try again."*

Sarina Behar Natkin

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# RESUMO

A locomoção é uma tarefa de grande importância na vida das pessoas. Ainda que pareça uma tarefa simples, andar é um exercício complexo que envolve controlo nervoso a fim de ativar os músculos e criar um movimento coordenado. Embora exista variabilidade natural nos padrões de marcha de indivíduos saudáveis, é possível definir um padrão "normal". O mínimo distúrbio a nível neuromuscular que afete a marcha de um individuo resulta na perturbação da qualidade de vida do mesmo, podendo mesmo condicionar a sua independência. Paralisia Cerebral, Esclerose Lateral Amiotrófica e Parkinson são algumas das doenças que podem afetar o padrão normal da marcha. Outra condição que pode desencadear alterações é o Acidente Vascular Cerebral (AVC), de acordo com a com a Organização Mundial de Saúde, cerca de 15 milhões de pessoas em cada ano sofrem um AVC, das quais 50% sofrem alterações da marcha não permanentes. Cada uma das condições mencionadas provoca alterações diferentes à marcha normal permitindo a definição de padrões de marcha de acordo com a condição que os afeta.

Por norma, o tratamento mais utilizado para distúrbios da marcha é reabilitação motora que consiste na realização repetida de exercícios que permitem a estimulação dos músculos de forma a que voltem a estar ativos. Ao longo do tempo as técnicas de reabilitação motora foram evoluindo e recentemente a engenharia uniu-se à medicina para originar uma nova área: a Reabilitação Robótica. Esta área faz uso de tecnologias robóticas com o objetivo de proporcionar um tratamento mais personalizado e adequado a cada paciente, beneficiando assim quer o paciente, quer os terapeutas. Embora ainda esteja em crescimento, esta área tem já demonstrado um grande potencial.

O Exoesqueleto é um dispositivo robótico que começou por ser usado em fins militares de forma a aumentar a capacidade que cada soldado carrega, é agora bastante utilizado na Reabilitação Robótica. Este dispositivo estimula o paciente a andar e vai apoiando conforme necessário, respondendo ao paradigma ajudar tanto quanto necessário, ou seja, o dispositivo ajuda o paciente a caminhar, dando-lhe apenas o impulso necessário para que este consiga prosseguir, tendo como objetivo final deixar de ser necessário enviar este impulso. Este procedimento é determinado pela estrutura de controlo do exosqueleto que consiste na estratégia que rege e define o comportamento do dispositivo robótico de acordo com a informação que os sensores do mesmo lhe fornecem. Por exemplo, existem controlos de posição, em que o exosqueleto conhece uma trajetória de padrão normal e ajusta a posição do paciente mediante a diferença que deteta entre a posição dita atual e a posição de referência.

A estratégia de controlo desempenha também um papel muito importante no âmbito da Reabilitação Robótica, é claro que os pacientes beneficiam de terapias o mais personalizadas possível, no entanto, o desenvolvimento de uma estratégia de controlo é um processo moroso e que envolve recursos. Uma possível solução para esta limitação é a simulação, que consiste na imitação de um processo ou sistema do mundo real em função do tempo, sendo usado para processos de otimização, testes, treinos e engenharia de segurança. Tendo isto em conta, simulação seria uma forma rápida e económica de estudar novas estratégias de controlo ou até otimizar já existentes.

O objetivo deste trabalho consistiu em desenvolver um modelo capaz de realizar simulações de um exosqueleto, mais especificamente do exosqueleto H1, desenvolvido ao abrigo do projeto HYPER. Este modelo foi desenvolvido em OpenSim, um simulador de uso livre desenvolvido pelo National Center for Simulation in Rehabilitation Research (NCSRR), Stanford University, USA. Este simulador é usado maioritariamente para projetos na área da biomecânica com especial enfoque para o estudo do comportamento de sistemas músculo-esqueléticos.

Primeiramente, foi efetuado um estudo intensivo sobre padrões de marcha, de forma a perceber quais as condições que podem afetar a marcha de um individuo. Este estudo apresenta a definição de alguns padrões de marcha como: (1) Padrão Normal, (2) Padrão Hemiplégico, causado por AVC, (3) Padrão Diplégico, causado por Paralisia Cerebral, (4)Padrão Neuropático, causado por Esclerose Lateral Amiotrófica, (5) Padrão Miotrófico, causado por Distrofia Muscular, (6)Padrão Parkinsoniano, causado pela doença de Parkinson. Além disto, foi realizada uma pesquisa bibliográfica de forma a conhecer o estado da arte das estratégias de controlo usadas na área de Reabilitation Robótica. Conhecer as características de um padrão de marcha, bem como dos controladores existentes é importante na medida em pode ser interessante desenvolver estratégias de controlo de acordo com o padrão de marcha ou pelo menos conhecer que padrões se devem ajustar para uma terapia mais eficaz de acordo com a condição que afeta o paciente.

A construção deste modelo iniciou-se no SolidWorks, um software de desenho assistido por computador, onde o sistema foi modelado de acordo com as propriedades físicas do H1, seguindo-se modelação por código em XML. Após a construção, o modelo foi validado. Para efetuar esta validação foram efetuadas provas estáticas e em movimento com o exosqueleto, tendo sido recolhidos os seguintes dados: ângulos e momento de cada articulação. Os momentos recolhidos nestas provas foram depois comparados com os momentos calculados com a ferramenta Inverse Dynamics do OpenSim, que usou como dados de entrada os ângulos de cada articulação.

O modelo construído, denominado *Exoskeleton*, foi depois integrado num novo modelo em conjunto com um modelo já disponível na base de dados OpenSim, o *3DGait2392*. A junção destes modelos deu origem ao *ExoBody*, um modelo que permite estudar a interação entre o dispositivo robótico e o paciente. Apesar de este modelo não ter passado por um processo de validação análogo ao do *Exoskeleton*, foi usado para um pequeno estudo de marcha onde se comparou a marcha de um individuo saudável com um paciente de AVC com e sem o uso do exosqueleto. Para a realização deste estudo foram utilizados data sets disponíveis online na base de dados OpenSim, estando já preparados para ser usados como dados de entrada das ferramentas Inverse Kinemaitcs e Inverse Dynamics. A Inverse Kynematics é uma ferramente que calcula para cada instante de tempo a posição do modelo que melhor corresponde à posição experimental, sendo esta determinada por marcadores por norma colocados na pele do individuo em estudo. A Inverse Dynamics, por sua vez, determina as forças generalizadas responsáveis por um determinado movimento em cada articulação.

Ambos os modelos construídos são capazes de realizar simulações no OpenSim sem gerar erros de sistema e dentro de tempos computacionais considerados normais. Tal como esperado, a comparação entre os dados experimentais e os dados simulados referentes ao modelo *Exoskeleton* foram concordantes e por isso o modelo foi validado com sucesso. Considerando o *ExoBody* model, os resultados apresentados evidenciam diferenças entre os padrões de marcha e também é possível verificar diferenças aquando do uso do exosqueleto ou sem o mesmo.

Posto isto, é possível concluir que os objetivos deste trabalho foram alcançados com sucesso uma vez que se desenvolveu o modelo que permite a simulação do exosqueleto bem como a sua personalização, adição de componentes como atuadores ou controladores. É importante referir que o modelo *Exoskeleton* tem algumas limitações, nomeadamente referentes ao design do mesmo que poderá ser melhorado.

Partindo deste trabalho, novos desafios podem ser enfrentados na perspetiva de continuar a melhorar e abrir horizontes na Reabilitação Robótica, nomeadamente, seria importante fazer uma validação do *ExoBody* incluindo um estudo de forças de reação.

**Palavras-Chave**: Distúrbios de marcha, Modelação, OpenSim, Reabilitção Robótica, Simulação.

# ABSTRACT

Locomotion plays a very important role in a person's life. Although healthy individuals show natural variability in gait patterns, it is possible to define an acceptable pattern for "normal gait". However, some pathologies as Amyotrophic Lateral Sclerosis (ALS), Spinal Cord Injury (SCI), Stroke or others can induce abnormal gait patterns that can limit the life of a person, making him/her dependent of others and consequently reducing his/hers quality of life. Robotics rehabilitation therapies are a growing solution that intends to revert or diminish the impairments in gait. The use of robotic devices, such as exoskeletons, cover some limitations of the traditional therapeutic methods, which is a great benefit for both patients and therapists. Furthermore, the application of an adequate treatment in these patients can be improved with the understanding of how the pathology affects the individual and through the development of specific solutions for each patient. Nowadays, computational dynamic simulations have great potential and help researchers to find optimal and personalized solutions for each patient.

Thus, the present work describes the development of an exoskeleton model in a neurorehabilitation perspective. First of all, a detailed description of gait patterns is presented, followed by the state of the art in robotics rehabilitation, considering that this field contains very powerful solutions for gait disorders. The model was developed in OpenSim, an open source software dedicated to model musculoskeletal systems and dynamic simulations of movement. In order to verify the accuracy of the model, experimental data were collected in static and motion trials performed with the wearable robot and afterwards compared with the simulated data resultant from Inverse Dynamics, a tool from OpenSim. The *Exoskeleton* model was successfully validated and then integrated in a new model, named *ExoBody*, within a musculoskeletal model. The *ExoBody* model was used to perform gait analysis comparing simulations with and without the exoskeleton, revealing some differences. Even though the built models present limitations, this work represents a step-forward in human-centered rehabilitation.


**Keywords:** Gait Disorders, Modelling, OpenSim, Robotics Rehabilitation, Simulation.

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| | |
|---|---|
| ALEX | Active Leg Exoskeleton |
| ALS | Amyotrophic Lateral Sclerosis |
| API | Application programming interface |
| AVC | Acidente Vascular Cerebral |
| CAD | Computer-aided design |
| CAE | Computer-aided engineering |
| CLME | Complementary Limb Motion Estimation |
| CMT | Charcot-Marie-Tooth |
| CP | Cerebral Palsy |
| CPG | Central Pattern Generators |
| CSIC | Consejo Superior de Investigaciones Cientificas |
| DMD | Duchenne Muscular Dystrophy |
| DOF | Degrees of freedom |
| GC | Gait Cycle |
| GUI | Graphical user interface |
| HAL | Hybrid Assistive Limb |
| NCSRR | National Center for Simulation in Rehabilitation Research |
| PD | Parkinson Disease |
| SCI | Spinal Cord Injury |
| SDK | Software development kit |
| SIMM | Software for Interactive Musculoskeletal Modeling |
| UPC | Universitat Politècnica de Catalunya |

# 1. INTRODUCTION

Walking is a trivial task for most of the people. One foot at a time and the subject moves forward. Although it seems very simple, gait is a complex task that involves nervous control to activate muscles and create a coordinated movement. The muscles are activated by electric impulses arriving through the neurons and sent from control mechanisms.

Since it is such a basic action, a minimal neuro-musculoskeletal disorder results in a perturbation to one's quality of life and independence, hindering several normal activities. Gait disorders are consequence of conditions such as Cerebral Palsy (CP), Amyotrophic Lateral Sclerosis (ALS), Spinal Cord Injury (SCI), Parkinson Disease (PD), or incidents such as Stroke. According to the World Heart Federation, every year, 15 million people worldwide suffer a stroke and it is the second leading cause of disability [1]. In 2013, the World Health Organization reported between 250 000 and 500 000 people affected by spinal cord injuries around the world, every year [2].

Gathering the best of both fields, medicine and engineering created robotics rehabilitation, an area that is showing a growing potential. The use of robotic devices, such as exoskeletons, cover some limitations of the traditional therapeutic methods, which is a great benefit for both patients and therapists. In spite of the big evolution in this area, there is still a lot to improve and explore.

Simulation is defined as the imitation of a real-world process or system over time, used for performance optimization, testing, training, or safety engineering. Considering this, simulation can be a very valuable tool in the development and improvement of medical rehabilitation devices.

The present work describes the modeling and simulation of an exoskeleton in OpenSim, a novelty in the neurorehabilitation field. This model allows researchers the possibility of exploring different conditions and variables leading to new control strategies or optimization of the existent ones, as well as a better understanding of the biomechanical behavior of the musculoskeletal system when in contact with the exoskeleton. Furthermore, the model developed can be a first step for a complete new strategy of planned and personalized rehabilitation. This document is organized in five chapters:

**Introduction** – Fundamental concepts to understand the work developed. First, a detailed description of Normal Gait is given including its characteristic Gait Cycle (GC) and respective phases and sub phases. Then, the characterization of several pathological gait patterns is presented as well as the state of the art of the solutions for the problem. The chapter ends with a brief description of simulation software and the definition of the objectives of the work.

**Methods** – The second chapter contains a detailed description of the implemented procedure including the software used in each step. First, it is explained the design of the exoskeleton, passing through the model code writing, and the testing and validation of the exoskeleton model. Second, the construction of a new model, *ExoBody*, integrating a musculoskeletal system and the simulation procedure is described.

**Results** – Here, the comparison of experimental data with simulated data is presented as result of the validation process. Furthermore, this chapter includes the results of a simulated gait analysis of a healthy subject and a stroke patient.

**Discussion** – This chapter focusses on the analysis and discussion of the results presented in the previous chapter. First, the validation results of the *Exoskeleton* model are discussed. Lastly, the chapter ends with gait analysis of two subjects with different gait patterns, using the *ExoBody* model.

**Conclusion -** As the final chapter, it presents the general conclusions of the present study, as well as the limitations of this study and possible future developments.

## 1.1 GAIT PATTERNS

Gait is a sequence of foot movements by which a human or animal moves forward [3]. Although every subject has its unique gait pattern, there are some common characteristics that allow for a normal gait pattern definition. Normal gait can be affected by several conditions as explained above and each of them can affect an individual in very different ways generating new gait patterns. Thus, this section presents a detailed description of Normal Gait and some pathological gait patterns.

### 1.1.1 Normal Gait

The GC is the period of time between any two identical events in the walking cycle and includes two phases: stance and swing, as can be seen in Figure 1.1. Stance is the time when the foot is in contact with the ground, representing approximately 60% of the cycle. Swing designates the time when the foot is in the air, and represents the remaining 40% [4]. Double stance is the period when both feet are in contact with the ground. It happens at the beginning and at the end of the stance phase, i.e., twice in a GC. Double support time decreases as walking velocity increases, for instance, there is no double stance in running. The period when only one foot is in contact with the ground is known as single support. The double stance period represents 10% of the GC (20% counting two events) while single stance accounts the remaining 40%. Additionally, the term *ipsilateral* is used to describe the same side of the body, and the term contralateral is used to describe the opposite side of the body or the opposite limb. Line of progression denotes the direction of walking [5].

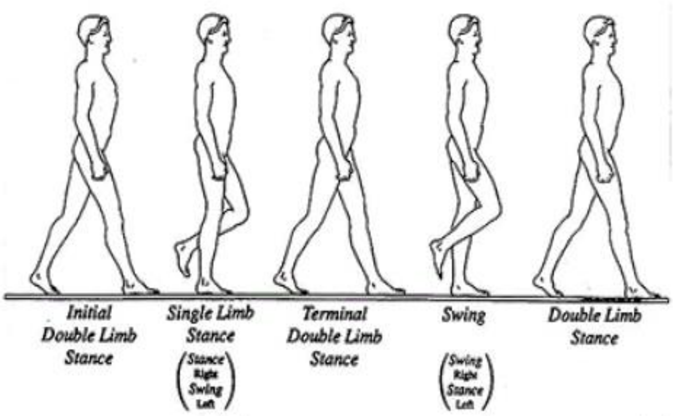



Figure 1.1 - GC includes two phases: stance and swing. Image from [4]

According to Ed Ayyappa [5], GC is responsible for three functional tasks: weight acceptance, single limb support and limb advancement.

Two phases of the stance period contribute to the weight acceptance: initial contact and loading response. Initial contact is the instant when the foot of the leading limb touches the ground, marking the beginning of the stance phase. Loading response consists in the period of initial double limb support. In this phase, a foot comes in full contact with the floor and the body weight is transferred to this limb. The immediate transfer of body weight onto the limb as soon as it contacts the ground requires initial limb stability and shock absorption while simultaneously preserving the momentum of progression. At initial contact, the external forces provoke a ground reaction that induces knee flexion. During loading response, eccentric quadriceps contractions allow the balance between knee stability and shock absorption. Achieving the weight acceptance means that the individual has a stable kinematic chain [5].

Single limb support is realized by midstance and terminal stance. Midstance constitutes the first half of a single support, it starts when the contralateral foot leaves the ground and continues as the body weight travels along the length of the foot until it is aligned over the forefoot. The terminal stance is the second half of single support and in this phase, the body weight moves ahead of the forefoot. During single limb support period, the contralateral foot is in the swing period, and total body weight is exclusively supported on the stance limb [5].

Limb advancement is characterized by its four phases: preswing, initial swing, midswing and terminal swing. The stance limb leaves the ground and advances forward to posture itself in preparation for the next initial contact. Preswing denotes the end of double limb support, it starts when the contralateral foot contacts the ground and ends with ipsilateral toe off. It marks the end of stance phase and the beginning of swing phase. In initial swing phase, the foot leaves the ground and continues until maximum knee flexion occurs, when the swinging extremity is directly under the body and directly opposite to the stance limb. Following this, midswing phase ends when the tibia is in a vertical position. Finally, in terminal swing, the tibia passes beyond perpendicular to the ground, and the knee fully extends in preparation for heel contact [5].

Gait characterization can be made using several parameters. Step length is the linear distance between 2 consecutive points of contact with the ground by the right and left feet. Stride length is the linear distance between 2 consecutive points of contact with the ground by the same foot, usually expressed in meters. Stride length comprises two step lengths, as can be observed in Figure 1.2. Stance time refers to the period in which the foot is in contact with the ground during the GC, and swing time refers to the period when the foot is not in contact with the ground during the GC [4], [5].

Velocity or speed is associated with both temporal and spatial gait data. Cadence refers to the number of steps taken per unit of time and is the rate at which a person walks, expressed in steps per minute [6].

Different gait patterns are characterized by differences in limb movement patterns, overall speed, forces, kinetic and potential energy cycles, and changes in the contact with the surface. Gait deviations from the typical pattern are often characteristic of specific neurological, muscular, or skeletal pathology.

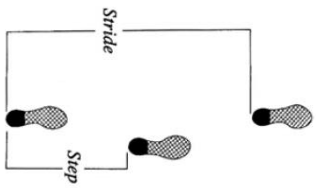



Figure 1.2 - Schematic representation of stride and step length. Image from [5].

### 1.1.2 Hemiplegic Gait

Stroke occurs when the blood reaching the brain is interrupted, cutting the provision of oxygen and nutrients and leading to damage in the brain tissue [7]. This problem is consequence of two possible causes, two types of stroke: **ischemic** – when a clot causes blood vessel obstruction, cutting the blood circulation -, and **hemorrhagic** – when a weakened blood vessel bursts, usually associated with high blood pressure. The effects of a stroke depend on which part of the brain is affected and how severely, some of the following functions, or all of them, may be impaired: movement and sensation, speech and language, eating and swallowing, vision, cognition, perception and orientation to surroundings, self-care ability or emotional control [7].

In particular, stroke can cause upper motor neuron syndrome which consequently results in an assemblage of sensorimotor impairments including muscle weakness, impaired selective motor control, spasticity, and proprioceptive deficits that interfere with normal gait [8]. Hemiplegic gait is the gait pattern commonly observed in stroke patients.

Diminished strength, or the inability to generate voluntary muscle contractions of normal magnitude in any muscle groups, and inappropriately timed or inappropriately graded muscle activity are the two immediate impairments of most significance to gait performance [9].

Normal gait tends to be symmetrical, both spatially and temporally, with interlimb differences in vertical force and temporal parameters measuring less than 6%. In contrast, hemiparetic gait is characterized by asymmetry, as it is represented in Figure 1.3, with poor selective motor control, delayed and disrupted equilibrium reactions, and reduced weight bearing on the paretic limb [8].

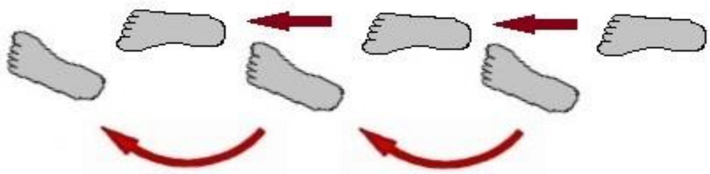


Figure 1.3- Schematic representation of hemiplegic gait. Image from [36].

In hemiplegic gait, temporal asymmetry is quantitatively described as a prolonged paretic swing time. However, the pattern of spatial asymmetry is less consistent since patients who have had a stroke constitute a heterogeneous group. The severity of the stroke, as well as the location and type, and the associated complications, determine to a large extent the level of dysfunction [8].

Hemiplegic gait is characterized by having a reduced walking speed, reported to range from 0.10 m/s to 0.76 m/s [8].The stance phase of both the affected and unaffected sides is longer in duration and occupies a greater portion of the full GC in subjects with stroke than in the healthy ones, walking at normal speed. Furthermore, comparing unaffected with affected side, the stance phase is longer in the unaffected limb [9].

Considering the kinematic perturbations, it is possible to analyze them separately, for stance and swing phases:

*Stance Phase* – Hip extension normally ranges from about 20º of flexion at initial contact to 10º of extension during the stance phase. The joint angles of a normal gait pattern are presented in Figure 1.4. In hemiplegic gait it is commonly reported that hip extension decreases [10].

The most common cause of a decrease in hip extension in the stance phase is over activation of the plantar flexor muscles because ankle dorsiflexion in the stance phase allows for forward progression of the leg, which is necessary for hip extension and advancing transport of the trunk segment to occur. Nonetheless, an increase in net plantar flexor moment, which could result from plantar flexor over activity or plantar flexor lack of eccentric contraction or shortening, limits ankle dorsiflexion and, consequently, hip extension in the late-stance phase [8].

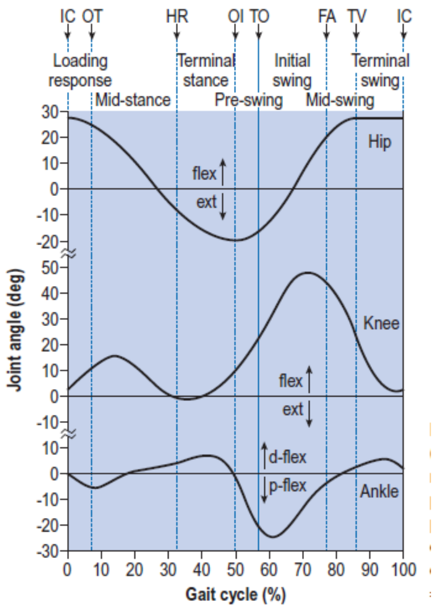



Figure 1.4 - Joint angles in a normal GC. Image from [41].

Three types of knee patterns during stance have been reported in persons with hemiplegic gait [8]:

(1) increased knee flexion (particularly at initial contact), relative to healthy control subjects;

(2) reduced knee flexion during the early-stance phase, followed by knee hyperextension in the late-stance phase and delayed movement into knee flexion in preparation for the swing phase;

(3) excessive knee hyperextension throughout most of the stance phase. When walking with a slow cadence, the knee flexes from about 3º at heel strike to 35º at toe-off in preparation for the swing phase.

The other common stance-phase kinematic disturbances are initial contact when the foot is flat and decreased plantar flexion at toe-off.

*Swing phase* - Two types of knee patterns are observed during the swing phase in hemiplegic patients. Normally, a peak knee flexion angle of about 65º is achieved during the first third of swing phase and 4º of flexion is reached immediately before initial contact. A decrease in knee flexion during the swing phase is among the most common gait disturbances in hemiplegic patients, and stiff knee gait is a characteristic pattern of this disturbance [8].

The second knee pattern in the swing phase is decreased knee extension prior to heel strike due to insufficient acceleration of the thigh or leg in the mid to late swing phase to produce the velocity-dependent moments that normally assist in knee extension. Also, due to excessive activation of the knee flexor muscles or a change in the length and stiffness of tissues on the flexor aspect of the knee, a lack of knee extension before heel striking might occur [8].

Another common swing-phase kinematic disturbance is decreased dorsiflexion of the ankle. Normally the ankle reaches a neutral position at midswing and maintains this position until initial contact. Ankle neutral position at this point in the GC facilitates limb clearance avoiding the toes from touching the ground.

In summary, swing-phase sagittal plane disturbances of hip, knee, and ankle motions in an hemiplegic limb are characterized by limited or reduced hip flexion, reduced knee flexion, and reduced ankle dorsiflexion or continuous plantar flexion. Limited hip and knee flexion and reduced ankle dorsiflexion increases leg length, which leads to a reduction in floor clearance by the foot during the swing phase, resulting in toe dragging or compensatory circumduction of the leg.

Researchers conclude that there is a deficit of bilateral symmetry in the kinetic patterns, with changes in both the involved and uninvolved lower limbs [8]. Kinetic analysis includes 3 variables: force, moment, and power.

The vertical ground reaction force may exhibit 2 peaks—one occurring at weight acceptance and the other at push-off and intermediate during midstance, as in healthy subjects. Some hemiplegic patients maintain a rather constant vertical force, with 3 or more small irregular peaks and troughs in the affected leg. The unaffected leg often exhibits a greater vertical force after initial foot contact and at push-off compared with the affected leg. The knee moments in hemiplegic patients differ from those in healthy subjects, who exhibit a negative flexor moment during early stance, followed by a positive extensor moment throughout the remainder of stance. Patients with hemiplegia exhibit a positive extensor moment throughout the GC in the affected lower extremity. There is an agreement that most moments and power bursts are reduced in amplitude in patients who have had a stroke and are smaller on the paretic side than on the nonparetic side and smaller in both limbs compared with healthy subjects walking at a comfortable gait speed. Gait speed in patients who have had a stroke varies from exceedingly slow to nearly normal, it is generally slow during the early stage of recovery and is associated with weakness and poor motor control of the lower extremity. As muscle strength and motor control improve, speed increases and abnormal movements decrease [8].

Hemiplegic gait can be resumed as: the patient stands with unilateral weakness on the affected side, arm flexed, adducted and internally rotated. Leg on same side is in extension with plantar flexion of the foot and toes. When walking, the patient holds his or her arm to one side and drags his or her affected leg in a semicircle (circumduction) due to weakness of distal muscles (foot drop) and extensor hypertonia in lower limb.

### 1.1.3 Diplegic Gait

The diplegic gait includes several types, however in this document, only one will be considered. In the diplegic pattern both limbs are affected. A patient walks with an abnormally narrow base, dragging both legs and scraping the toes as represented in Figure 1.5. This gait is a consequence of bilateral periventricular lesions, such as those seen in Cerebral Palsy (CP) [11]. These brain lesions occur during brain development before or after birth and given the complexity of this process the results of an injury or abnormal development are varied leading to different presentations of CP [12].

Figure 1.5 – Both limbs are affected in diplegic gait pattern. Image from [52]

CP is the most common physical disability in childhood. Motor disorders of children with CP are related to primary deficits, such as, spasticity, muscle weakness, reduced coordination, and a loss of selective motor control. Also, secondary deficits, such as muscle contracture and bone deformities [13].

As the CP patient grows, his locomotion skills deteriorate mostly because of pain, fatigue, and shortage of adapted physical activity. Therefore, the gait pattern in adults is not very well documented.

However, there are some studies of patterns in children. In [13] is presented a research work which concludes that spatiotemporal gait parameters, including walking speed, cadence, and stride length are significantly lower in children with spastic diplegic CP than in non-disabled children. Step width is larger in children with CP due to poor balance and gait instability. Moreover, significant differences in proportions of the stance phase (single- and double-limb stance) are presented in children with spastic diplegic CP compared to healthy children. The duration of single-limb support in children with spastic diplegic CP is shorter, and consequently double-limb support is longer.

According to [14], maximal hip extension, hip range, knee range, knee flexion, time of peak knee flexion, and maximal ankle dorsiflexion in stance and swing phases are the gait parameters that characterize a CP gait.

### 1.1.4 Neuropathic Gait

This kind of gait is observed in patients with weakness of foot dorsiflexion, commonly known as foot drop. It happens due to an attempt to lift the leg high enough during walking, so that the foot does not drag on the floor. Some of the causes are Charcot-Marie-Tooth disease (CMT), Amyotrophic Lateral Sclerosis (ALS) and other peripheral neuropathies including those associated with uncontrolled diabetes [11].

CMT is a group of inherited disorders that affect the peripheral nerves - the nerves external to the brain and spinal cord. It has a prevalence of 1 case in every 2500 and clinical manifestations involve the peripheral nervous system and are generally characterized by symmetrical weakness and atrophy of distal muscles, predominantly in the lower limbs [15].

In CMT, gait pattern is characterized by a decreased capacity to raise the foot from the floor during the swing phase, with compensation achieved by increased knee and hip flexion. However, this ‘‘steppage gait’’ description cannot reflect the complex interaction between muscle deficit, structural alterations, biomechanical dysfunctions, and compensatory adjustments occurring during the course of the disease. In addition, patients affected by gait disturbances may not only have foot drop, but also a plantar flexor deficit, which could further impair gait [15].

In [15] several patients were analyzed and compared with healthy individuals, revealing that CMT patients displayed a significantly longer stride duration, lower swing velocity, shorter step length and greater step width than controls, although no differences were found between patients and controls in the percentage of swing duration. The main differences between CMT patients and controls were found in the kinematic and kinetic behavior of the ankle, with a detrimental effect on the walking ability as shown by a decrease of step length, cadence, and swing velocity.

ALS is a neurological pathology that affects the neurons responsible for controlling voluntary muscle movement. The muscle fibers are denervated as their corresponding anterior horn cells degenerate. Also, the spinal cord’s lateral columns and the upper motor neurons’ axons degenerate and are replaced by fibrous astrocytes [12]. The disease is progressive and currently there is no cure or effective treatment to halt, or reverse the progression of the disease.

A study [16] concluded that, compared to healthy subjects, people who suffer from ALS have longer stride time and slower walking speed. The gait is characterized for being less steady and disorganized. These alterations happen due to muscle weakness, decrease endurance and muscle fatigability, which are consequences of ALS.

### 1.1.5 Myopathic Gait

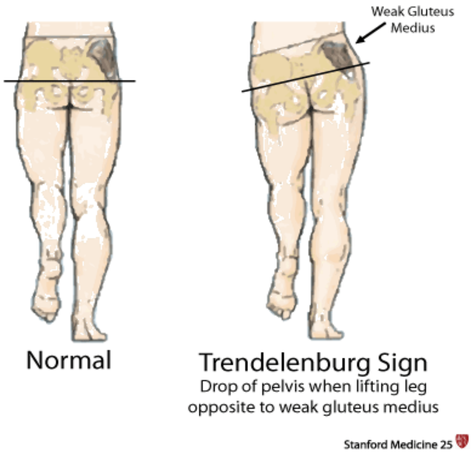



Figure 1.6 - Difference between normal and myopathic gait. Image from [40]

Hip girdle muscles maintain the pelvis level when walking. If there is a weakness on one side, this will lead to a drop in the pelvis on the contralateral side of the pelvis while walking (Trendelenburg sign), as represented in Figure 1.6. With bilateral weakness, there will be a dropping of the pelvis on both sides during walking leading to waddling. This gait is seen in patient with myopathies, such as muscular dystrophy [11].

Muscular dystrophy is a progressive genetic disorder that causes deterioration of the muscles and eventually leads to muscle wasting, muscle weakness, bone deformities and disability. There are many different types of muscular dystrophy. All types cause muscle weakness, but the areas affected and the severity of the symptoms can differ.

Similarly to CP, patients of muscular dystrophy lose locomotion skills with aging. Hence, D'Angelo et al. [17] concluded from a study of 21 patients with Duchenne Muscular Dystrophy (DMD), comparing them with 10 healthy controls, that DMD patients have knee hypertension and excessive abduction in swing phase at the hip level on frontal plane to aid clearance. Also, hip and ankle power were significantly reduced in DMD children. Velocity and cadence in DMD patients was similar to those in healthy subjects, however, stride length was reduced and step width was increased to improve balance. A similar but more recent study [18] confirms those conclusions.

### 1.1.6 Parkinsonian Gait

Parkinson Disease (PD) is a disorder of the basal ganglia, which is a group of nuclei situated at the base of the forebrain, responsible for modulating the cortical output necessary for normal movement [12]. Dopamine is a neurotransmitter, produced in basal ganglia, that has an important role in motor control and controlling the release of several hormones. PD appears when the dopamine-production cells die leading to decrease levels of the neurotransmitter causing PD symptoms to appear [19].

The Parkinsonian gait is the gait pattern of PD patients and it is characterized with rigidity and bradykinesia. An individual stands with the head and neck forward, with flexion at the knees. The whole upper extremity is also in flexion with the fingers usually extended. The patient walks with slow little steps known as *marche a petits pas* (walk of little steps). Figure 1.7 presents a schematic representation of the main characteristics of Parkinsonian gait. Patients may also have difficulty initiating steps and it's common to show an involuntary inclination to accelerate steps, known as festination. This gait is seen in Parkinson's disease or any other condition causing parkinsonism, such as side effects from drugs [11].


Figure 1.7 - Schematic representation of Parkinsonian gait. image from [21].

Peng Ren et al., [20] compared Parkinson's disease (PD) patients with healthy subjects and found differences in the GC: for normal gait, the heel strikes the ground before the toes, which is known as heel-to-toe walking. However, in Parkinsonian gait, motion is characterized by flat foot strike or the toe-to-heel pattern. Therefore, the foot strike patterns (or sub-phases of stance) of PD patients are totally different from that of healthy people.

In [21] it is related that PD patients tend to have shortened stride length, reduced overall velocity and increased stance phase durations. In addition, reduced or absent arm swing, reduced trunk rotation and decreased amplitude of motion at the hips, knees, and ankles are also reported characteristics of the PD gait. After studying several PD patients off medication or therapy, the authors characterized Parkinsonian gait with short steps, lower walking velocity and relatively high cadence than control subjects. Regarding joint angles, they report deficit of hip extension, decreased knee extension during single stance support, and reduced plantar flexion of the ankle at the toe-off. Concerning ankle joint kinetics, they found reduction of the ankle dorsal-flexion moment during loading, reduction of the maximum ankle extension moment and of the ankle power generated in the preswing phase of the GC (push-off power). At the knee joint, there is a lower power generation during the single GC stance phase, which occurs with a limited knee extension during the stance phase of gait. PD patients did not generate sufficient power to extend and thereby passively stabilize the knee as the control subjects did. Decreased power absorption at the knee joint during the late stance and preswing phase could be an effect of the impaired push-off power generation at the ankle. Lastly, for the hip kinetics of PD patients, it was reported a reduction of the maximum extensor moment and maximum flexor moment while power generation is decreased in double support and preswing [22].

## 1.2 STATE OF THE ART IN ROBOTICS REHABILITATION

The presented disorders are becoming common and seriously limiting the ability of elderly and patients to achieve their daily activities. Current research challenges refer to the development of new therapeutic methods and assistance modes that help impaired people to improve their daily life activity performances and to restore lost or impaired motion control.

Orthopedic rehabilitation usually involves execution of specific movements to provoke motor plasticity and ultimately improve motor recovery. It is crucial for patients to improve their musculoskeletal strength and motor control and to minimize functional deficits.

Robotic solutions are growing and showing that they can cover some limitations of the traditional therapeutic strategies. For instance, physiotherapy is physically demanding for both therapist and patient. With robotic exoskeletons the patient might have more intensive training, and it gives a better quantitative feedback [10][23].

This section presents the state of the art of lower limb exoskeletons through a characterization of the mainly used control strategies. Although there are several Load Carrying Augmentation exoskeletons they will not be referred since they are not a valid solution for the problem.

The control strategy is designed based on the principle that the robots won't hinder human motion and that the robot can act according to the humans' intention. The main goal is to find the most effective control algorithm in order to produce greater rehabilitation benefits as well as ensuring the safety of the user. Furthermore, all the rehabilitation robots try to accomplish the *assist-as-need* paradigm, that consists in the principle of helping patients perform a movement with the minimal amount of external assistance possible [24] [25].

For a personalized therapy different control algorithms or an adjustment of the control parameters given a control strategy are necessary, i.e., according to a patient gait disorder and consequent gait pattern the best therapeutic strategy is different. Therefore, it is important to adapt control strategies for different conditions. A simulator can provide realistic imitation of the controls and operation of a system. Simulate a control strategy before implementing it on the robotic device is very advantageous since it gives the insight if one control strategy is really good or needs optimization, thus reducing time and resources.

According to [26], control algorithms can be classified considering the strategy used to provoke plasticity: assistive or resistive. Furthermore, there are more controllers presented in the literature such as the oscillator-based control here presented.

### 1.2.1 Assistive controllers

The simplest control method is position control [27] [28], by controlling each joint to track predefined trajectories. However, this control method reduces the adaptability of the system.

This control strategy is especially useful when the user has little ability to interact with or control the exoskeleton. In a predefined gait trajectory control mechanism, gait data of a healthy person is recorded and then relayed on an exoskeleton. However, it tends to give the user less control and interaction with the device. State machines can help address this problem and are employed in some of the designs to incorporate a combination of position and force control [29]. Due to the transitional nature of the GC (particularly swing and stance phases), it is often useful to break up the controller into multiple different control states depending on the phase of the GC. State machines can also be further extended to allow for different states for alternate terrain, such as stairs or transfers between standing and sitting among other activities.

Generally, there are two methods to define the trajectories used for position control:

1. Predefine a trajectory consistent with normal gait. The predefined trajectory includes joint angles derived from mathematical models of normative gait trajectories or imitated from prerecorded gait trajectories of healthy subjects
2. Define an instant trajectory from the unimpaired lower limb. Motions of the unimpaired lower limb are detected and dynamically mapped to the reference trajectory of the other limb using Complementary Limb Motion Estimation (CLME) or Central Pattern Generators (CPG).

The impedance control [30] [31] is an extension of position control and it does not only control the position and the force but also controls a relation and an interaction between the exoskeleton and the human body. The impedance model receives the error position of the joints and generates the force values that become the force references for the next stage, the force/torque controller. The force controller will try to guarantee that the forces exerted by the exoskeleton are equal or close to the force references.

Additionally, a variant of impedance-based assistance is a triggered assistance that works based on the impedance assistance but initiates after some performance variable reaches a threshold. This variable could be elapsed time, force generated by the patient, limb velocity or muscle activity [30][32].

Hybrid Assistive Limb (HAL) [33] [34] is an exoskeleton that uses position based control. It has an autonomous controller assisting the hip and knee joint while the ankle joint behaves passively as a spring. The hip and knee joints are controlled based on two phases: swing and stance phase. The desired joints patterns are pre-recorded from a healthy subject and are allocated to these two phases by a real-time intention estimator (understanding when the user starts walking, stops walking or has a leg swing) which counts on floor reaction force and torso angle.

Lokomat [33][34] is developed to support patients who have difficulties in lower-limb functions of sitting, walking, and standing. Likely HAL, hip and knee are actuated, while ankle is passively actuated with a spring. A finite state machine is used to determine the movements of the two exoskeleton legs. During walking, one stride cycle is separated into four states: left swing, left double stance, right swing, and right double stance. Stance to swing transition is triggered by the user moving his/her crutches and shifting his/her body weight. Swing to stance transition is triggered by heel strike detection.

Vanderbilt [29] was developed in Vanderbilt University and it works with an assistance controller, but it does not provide a desired trajectory or angle joints. This exoskeleton high level control uses a finite state machine, which sends the status information to joint controller about phases in gait. The controller was developed combining three types of behaviors: gravity compensation, feedforward movement assistance during swing and knee joint stability reinforcement during stance.

### 1.2.2 Resistive Controller

It is documented that physical guiding may decrease motor learning [19] [25]. Resistive controllers are an alternative to assistance controllers, working in the opposite way: they make a task harder to realize, challenging the patient. This therapeutic strategy offers resistance to the hemiparetic limb movements during an exercise in order to force that limb into making a bigger effort.

Another strategy used by resistive controllers is the constraint-induced strategy where the unimpaired limb is constrained to encourage the use of the impaired limb.

These therapeutic strategies are only used in lower level impairments.

Active Leg Exoskeleton (ALEX) [36] is a commercialized exoskeleton that has a force-field controller that applies desired force fields on the moving leg. In this case the controller can be assistive or resistive and the user is not restricted to a fixed repetitive trajectory.

### 1.2.3 Oscillator-based control

This method is based on adaptive oscillators, i.e., mathematical tools that are capable of learning the high-level features (frequency, envelope, etc.) of a periodic input signal. It is trajectory-free, in the sense that it provides the user assistance, regardless of the performed movement, requiring no sensing other than the assisting robot’s own encoders [37].

Oscillator control uses no other sensing than the encoder of the robot actuators, avoiding the problems related to sensor placement, user-dependent calibration, or signal durability and reliability. Therefore, this method provides both a fast and convenient integration to the user’s body and an adaptability to the user’s intentions which—pending a sound and attractive ergonomic design—are the major requirements to maximize the device acceptability for potential users.

Instead of directly estimating the intended movement kinematics (the epiphenomenon of the intended movement), this approach uses adaptive oscillators and uses the a*priori* knowledge that the movement is periodic to derive a non-linear dynamical system able to represent the movement in a finite set of simple features.

LOPES [34] is a frame-based treadmill mounted exoskeleton that works with adaptive oscillators. The control is realized in a model-free mode: in real-time operation, a pool of adaptive oscillators are adopted to extract the phase and frequency of hip joint angle. Then the phase and hip joint angle are fed to a kernel filter estimating the predicted hip joint angle without delay. The desired joint torque is computed as to attract the hip joint to its predicted angular position, by multiplying the difference between predicted and current hip joint angles with a virtual stiffness. This assistive strategy requires no extra sensors except the encoders, already integrated in the exoskeleton.

ALEX II [38] is an assistive device focused on the hip flexion and extension. The control works by exerting a hip torque based on online gait analysis results, with the current percentage of GC furnished by the adaptive oscillators: the gait stride initiation is detected by a foot-pressure, then an adaptive frequency oscillator is utilized to track the hip joint angle signal and extract its periodicity features, detecting also the instantaneous walking speed. By combining these two pieces of information, the instantaneous stride cycle percent is computed out.

A significant issue that remains present is how to effectively control the exoskeletons in order to maximize the benefits of these robotic devices. Controllers are very different from one exoskeleton to another, and few studies test different controllers directly on the same hardware. Additionally, most exoskeletons only have mechanical sensors embedded in the device. Even though these are useful, additional information about the state of the user state may prove to be very valuable for strengthening the control options that the user has over the device.

One of the most direct ways to incorporate user input and feedback is through a neural (or muscular) link. The muscular activity gives a representation of what the user is attempting to do and can be used to proportionally determine the torque generated by actuators in the device. Even though surface electromyography (EMG) is useful, its stability is easily compromised by several factors such as placement, fatigue, and sweat. Because of that, the use of surface EMG needs frequent recalibration, limiting the practicality of this technic. An alternative solution that holds promise for EMG-based technology is the use of intramuscular EMG electrodes that may have more stable properties over time. Despite muscle fatigue might still change signal properties over time, other problems as placement, sweat and electrode shift during movement should be solved. This technology could be highly beneficial for the subset of exoskeleton users who have both remaining EMG activity and permanent access to an exoskeleton device [28].

For paraplegics or people with severe lower-limb disability, a predefined trajectory control is mostly used. The trajectory is recorded from a healthy person with specialized material for the effect, or extracted from clinical gait analysis data, and then processed and replayed. To improve flexibility, comfort and security, more recent studies suggest parameterizing the joints trajectories in accordance with the users' body conditions and movement phases.

## 1.3 GAIT ANALYSIS AND SIMULATION

Gait analysis plays an important role in clinical decision and treatment design. It embraces three main topics: kinematics - temporal and spatial study of a movement -, dynamics – study of the forces involved in the movement -, and muscle activity. With the meaningful information provided by these tools it is possible to know one's gait pattern, the involved forces in articulations and each muscle actuation. Hence, the treatment design results from the combination of these information and the available solutions.

Health care simulation pretends to mimic aspects of clinical care through computer/screen-based simulators, task trainers, mannequins, or other devices. Simulation software can perform test hypotheses, predict functional outcome, and identify emergent behaviors. These features make modelling and simulation a very powerful tool to solve complex engineering problems, having the potential to revolutionize medical decision making and treatment design.

SIMPACK, AnyBody Modeling System, MADYMO, LifeModeler, Software for Interactive Musculoskeletal Modeling (SIMM) and OpenSim are simulation software dedicated to modelling and biomechanics analysis. These software use multibody analysis methodology to carry out kinematic and dynamic studies allowing any user to control numerous variables.

In this section, Gait Analysis and OpenSim will be explained in more detail.

### 1.3.1 Gait Analysis

Michael Whittle defined gait analysis as "the systematic study of human walking, using the eye and brain of experienced observers, augmented by instrumentation for measuring body movements, body mechanics and the activity of the muscles."[39]. The use of gait analysis has increased in the last decades, not only in the clinical field, but also in sports and research.

Gait analysis is based on information collected with technological tools in an appropriate environment. For instance, a visual gait analysis, that consists in unaided observation of the patient walking by one or more specialists, is considered a semi-objective analysis, since it is highly dependent on the specialist evaluation [40]. This kind of assessment allows for a greater knowledge of general gait parameters as cadence, cycle time, speed, or step length.

Kinematic analysis observes and describes the body's movements not considering what causes each movement, but focusing on the measurement of position and orientation of body segments, joint angles and corresponding velocities and accelerations. This analysis usually uses high-speed cameras placed in different points in the working volume making possible to obtain a 3D analysis. LED (light-emitting-diodes) can be used with an optoelectronic camera able to determine the position of each marker by examining the light that comes from it. With the position marker data it is possible to calculate joint angles and velocities [41].

Dynamic analysis focusses on joint forces, moments and powers. Power generation is a measure of the rate of work generated by a muscle or muscle group and it is useful because it allows to know whether muscles are contracting concentrically or eccentrically. Dynamics can also study the ground reaction forces (GRF) – forces that oppose to those applied by the foot on the floor, having an opposite direction and equal magnitude. Dynamic analysis requires kinematic data and external forces measurements usually obtained with force platforms [41].

The depolarization of a muscle membrane originates an electrical signal that consequently causes muscle contraction. EMG uses electrodes that measure the electrical signals and in that way register the muscle or muscle group activity [42].

### 1.3.2 OpenSim

OpenSim is an open-source software developed by the National Center for Simulation in Rehabilitation Research (NCSRR), Stanford University, USA, that allows building, changing, and analyzing musculoskeletal models and dynamic simulations of movement. The first version of OpenSim was released in 2007 and the OpenSim 3.3 in 2015. This framework includes an end-user application with a graphical user interface (GUI), a set of command-line utilities, and a software development kit (SDK) including application programming interfaces (APIs). The software is written in ANSI C++, and the GUI is written in Java, allowing OpenSim to compile and run on common operating systems.

OpenSim was developed aiming to encourage the biomechanics community to build a library of simulations that can be exchanged, tested, and improved through multi-institutional collaboration. Thus, after 10 years of the first version release, there is available online a library of musculoskeletal models of the whole bone or isolated parts, which the users can obtain, adapt, and improve.

Besides modelling a musculoskeletal system, OpenSim has useful tools that can perform on the gait analysis.

It is possible to scale an existent model to match patient-specific measurements, and then simulate the scaled model through: **Inverse Kinematics tool -** determine internal coordinates that best reproduce the experimental position of markers corresponding to known landmarks on rigid segment; **Inverse Dynamics -** that calculates the generalized forces necessary to achieve the model desired kinematics according to the basic equations of motion; and, **Forward Dynamics tool -** that finds the solution of the system dynamic differential equations.

Furthermore, OpenSim allows the user to define several variables that control muscles activities making possible the simulation of different conditions.

## 1.4 MOTIVATION

The World's population is increasing and its habits are changing with direct impact in public health. Although life expectancy increases, or several diseases have been eradicated, nowadays we face new challenges.

For instance, eating habits have changed, obesity is now common, causing high blood pressure which is usually associated with high levels of cholesterol, which may cause vessel obstruction leading to stroke. Unfortunately, like stroke, there are a lot of conditions that at the present time we are used to heard of, or even knowing someone who suffers from them.

As presented in the first section of this chapter, gait disorders have several causes, but despite the condition, the problem is real: a person suffering from a gait disorder has less life quality and might even depend on others to perform a normal life.

Even though there are some solutions available, they just achieve a potential alleviation of the damage, and thus, it is necessary to go even further, to improve, and find more effective solutions. Robotics rehabilitation is growing and proving results in Neurorehabilitation, aiming to achieve an effective patient recovery.

With that being said, the main objective of this work is to advance a step-forward in human-centered rehabilitation approaches through modeling and simulation of an exoskeleton in a robotics rehabilitation perspective. The literature research revealed that there isn't any open source model of an exoskeleton. Therefore, the existence of such model would cause impact in robotics rehabilitation since it enables the study of the exoskeleton behavior depending on different variables, such as weight, applied forces, motors, or others. Furthermore, it is known that a rehabilitation therapy should be as personalized as possible, so, the model is a powerful tool in the development of new control strategies allowing to adapt the control strategy to each gait pattern or even to each patient according to its characteristics.

The work presented in this dissertation was developed during a 9 months internship in the Robotics Laboratory of the Universitat Politècnica de Catalunya (UPC) under the supervision of Dr. Alicia Casals and Dr. Hugo Alexandre Ferreira.

The Robotics Laboratory is the workplace of the Research Group on Intelligent Robots and Systems (GRINS), which among other areas, works in rehabilitation robotics, focusing on development of control strategies for assist as needed rehabilitation. GRINS has a wearable exoskeleton H1, that was developed as part of Hybrid Neuroprosthetic and Neurorobotic Devices for Functional Compensation and Rehabilitation of Motor disorders (HYPER) project. As a matter of fact, the background of the group and their facilities were a great advantage for development of this work.

# 2. METHODS

Gait training is the most used solution for gait disorders. In the last few years, this solution has been revolutionized with the introduction of wearable robotic devices. Moreover, patients benefit from personalized therapy, adapted to their conditions and necessities. Thus, with this in mind, it was created an OpenSim model of an exoskeleton, more specifically of H1. The model denominated *Exoskeleton* aims to simulate possible improvements to do to the real exoskeleton. Furthermore, this model was integrated within a musculoskeletal system model, generating a new model, the *ExoBody*, in order to enable the study of the interaction between the robot and patient and new control strategies.

Chapter 2 focusses on the methods used in this task. First, the steps of the *Exoskeleton* model construction are explained in detail, followed by a description of the validation process. Second, the integration of the *Exoskeleton* model within the *3DGait2392* model is described. Lastly, the procedure to realize a gait analysis in OpenSim is presented.

As explained above, OpenSim is an open source software used for developing musculoskeletal models and to perform simulations of dynamic healthy and pathological movement. The software features can be easily accessed via a GUI, via instruction commands using command prompts or via the MATLAB interface, allowing batch processing of the analyses. Furthermore, OpenSim has an application programming interface (API) where users can implement new functionalities to the software.

Although there are several OpenSim musculoskeletal models published, allowing for a complete study of human motion, there is not a lot of work done in modelling and simulation of rehabilitation robotics. The development of a musculoskeletal model including an exoskeleton enables the study of the interaction between patient and device and how could this relationship be improved.

## 2.1 EXOSKELETON H1

The exoskeleton H1, presented in Figure 2.1.a, was developed and built in the Neural Rehabilitation group, Consejo Superior de Investigaciones Cientificas (CSIC), Madrid, as part of HYPER project [43]. The robotic device works as a rehabilitation tool for adults in the range of 1.50m to 1.90m height and with a maximum body weight of 100kg, with pathological gait.

H1 weights about 9kg, built in aluminum and stainless steel for mechanical resistance and lightweight. Hip, knee, and ankle in each leg are powered joints, translating in its 6 degrees of freedom (DOF). Each joint has a harmonic drive brushless DC motor enabling active and passive movements along the sagittal plane. The maximum achievable joint angles and torque values are presented in Table 2.1 [44].

Table 2.1 - Mechanical limits of the exoskeleton H1 [44]. The table presents the Joint Angles in degrees and the Motor Torques in Newton Meter, for each joint. The values are equal for right and left joints.

| Joints | Joint Angles (deg) | Motor Torques (Nm) |
|---|---|---|
| ***Hip*** | -20 to 100 | ±40 |
| ***Knee*** | -5 to 100 | ±40 |
| ***Ankle*** | -15 to 20 | ±20 |

The wearable device contains kinematic (angular position, velocity and acceleration) and kinetic (force of interaction between limb and exoskeleton) sensors, and strain gauges are used as force sensors designed to measure the torque produced by the interaction between the user and the device. Moreover, there are two force sensors in the footplates, that detect foot contact between foot and the ground, aiming to detect different phases of GC [44].

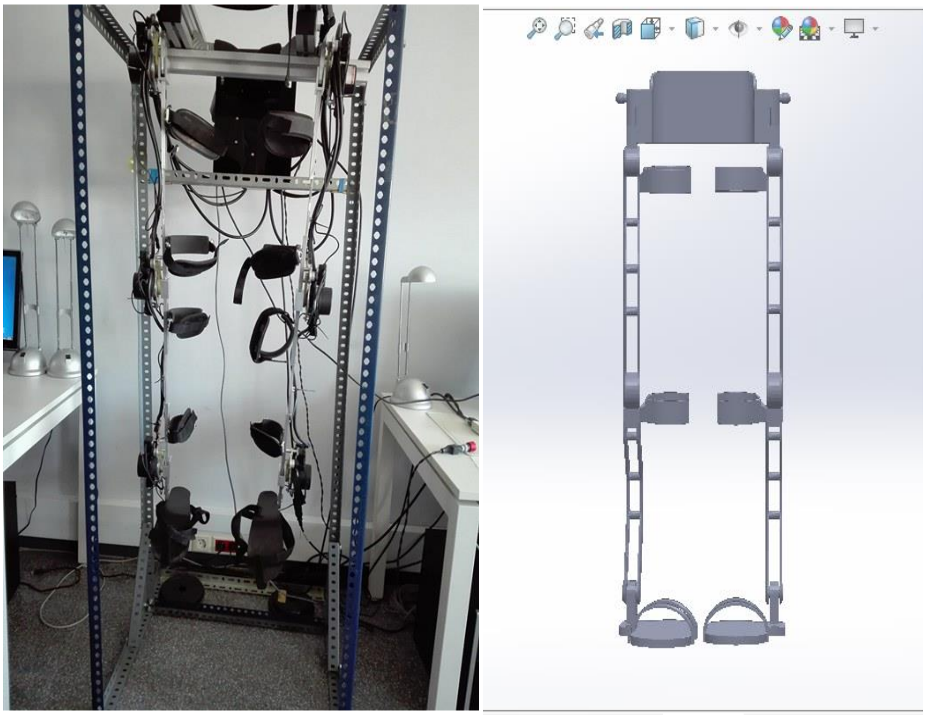
## 2.2 MODELLING PROCESS

First of all, it is important to introduce a few basic concepts of the OpenSim workflow. In general, OpenSim models represent dynamic systems of rigid bodies connected through joints that are acted upon by forces producing a motion. In the OpenSim environment the term "*body*" usually defines a rigid body, for instance, a bone segment, and "*joints"* denotes the coordinates and kinematic transforms that control the motion of that body with respect to its parent body. All *bodies* are connected to a parent via a *joint*, except for ground. Hence, the first step to define a model is to define a set of rigid bodies, and to do so it is necessary to know its specifications.

SolidWorks is a computer-aided design (CAD) and computer-aided engineering (CAE) software. Using this software is possible to develop CAD models that virtually represent something with its real characteristics. In this specific case, SolidWorks was used in order to model the exoskeleton design since it creates geometry files compatible with OpenSim requirements. In addition to that, SolidWorks is also a useful tool for calculating the mass, center of mass and inertia matrix of an element according to its physical properties.

The first step in the development of the *Exoskeleton* model was having a file with the geometry of an exoskeleton with the characteristics of H1. Since there was no specific CAD model of H1, a model from the GrabCAD Comunity [45] was used. This model was an assembly file, meaning that it has several parts that make a body. However, OpenSim only accepts .stl , .obj and .vtk files. Thus, the assembly file was edited in SolidWorks so it could be adjusted to H1 parameters, as weight and height, and compatible with OpenSim software.

The model was scaled proportionally to H1 and the proper materials were assigned to each part: *stainless steel* for all parts except the feet which were *rubber*. According with the material defined, SolidWorks calculates the mass and inertia matrix of the body. Furthermore, the CAD model must have the same Coordinate System than OpenSim. To adjust the origin of the assembly file it was necessary to create a new coordinates system and then, when saving the file choose the coordinates system wanted and the format file pretended. In this case, every part was saved as an .stl file with a Coordinate System equal to the OpenSim one. Figure 2.1 presents on the left side a photography of the real exoskeleton and on the right side a screen capture of the SolidWorks environment with the designed exoskeleton.



(a) (b)

Figure 2.1- (a) Lower limb rehabilitation exoskeleton H1, in the Robotics Laboratory. Force sensors at each link measure the torque between user's limb and robot. The exoskeleton is suspended in the air, fixed to its support. (b) Exoskeleton model on SolidWorks environment, with similar design and characteristics to H1 exoskeleton.

Thereby, the exoskeleton design was divided into seven parts: BackExo, Hip, Knee and Foot, being the last three for each limb. Consequently, in the model code seven bodies were defined. OpenSim models can be created through the API, using C++ or through a text editor using XML. Due to the existence of more support and examples, the model was written in XML using Notepad ++ as text editor.

The Mass Properties tool of SolidWorks was used to calculate each body mass, mass center and inertia matrix, as it is possible to see in Figure 2.2, where the tool calculated the properties for Right Hip body. Afterwards, the information given by this tool was used in the code to settle a body in the model as can be seen in Figure 2.3 where a code segment defining the right hip is presented.

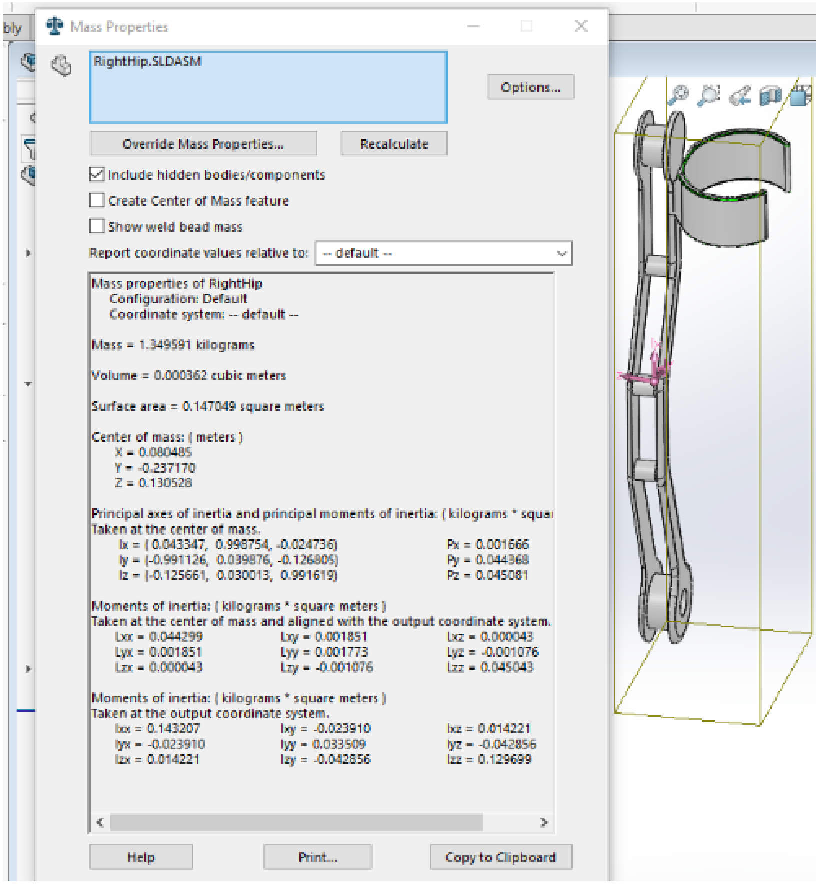



Figure 2.2 - Screen capture of Mass Properties tool in SolidWorks, this tool presents the information of mass center, mass and inertia matrix in International System Units.

```xml
<BodySet>
    <objects>
        <Body name="ground">
        <Body name="BackExo">
        <Body name="Hip_R">
            <mass>1.349591</mass>
            <mass_center> 0.080485 -0.237170 0.130528</mass_center>
            <inertia_xx>0.143207</inertia_xx>
            <inertia_yy>0.033509</inertia_yy>
            <inertia_zz>0.129699</inertia_zz>
            <inertia_xy>-0.023910</inertia_xy>
            <inertia_xz>0.014221</inertia_xz>
            <inertia_yz>-0.042856</inertia_yz>
```

Figure 2.3 - Code segment, in Notepad ++ environment, of the model defining the right hip body. The information present in the code is given by the Mass Properties tool presented in Figure 2.2.

As explained before, all bodies need a parent and it is necessary to define the kinematic relationship between them, i.e., to define a joint. OpenSim categorizes joints in seven types [46]:

- Weld / Rigid joint – introduces no coordinates and fuses bodies together;
- Pin / Revolute joint – one coordinate turns around the common Z-axis of parent and child joint frames;
- Slider / translational joint – one coordinate along common X-axis of parent and child joint frames;
- Ball / Spherical joint – 3 rotational coordinates that rotates about x, y, z of body and parent;

- Ellipsoid joint – 3 rotational coordinates that turns about x, y, z of body in parent with coupled translation such that body traces an ellipsoid centered with respect to the parent link;
- Free joint – six coordinates with 3 rotations and 3 translations of body with respect to the parent link;
- Custom joint – user specifies 1-6 coordinates and user defines spatial transform to locate B with respect to P.

With these different types in mind, Pin / Revolute Joint was the joint type chosen to connect all the parts of the model. This decision was based on two factors: (1) The exoskeleton only moves in the sagittal plane, (2) there is no need of using complex elements, as it would increment the computation time of the model and that was not intended.

Therefore, to characterize a Pin Joint, it was necessary to define the parent body, the joint location and orientation of the referred body and its parent. To define the joint location it is necessary to define the position and orientation of the joint in the body coordinates system and in the parent body coordinates system. Besides that, the definition needs to include the *Coordinate*, which concerns to the definition of the motion type and the range of values that the movement can range. For instance, Figure 2.4 presents the code segment with the definition of the joint and its parameters relative to the *Right Hip* body, that is connected to the *BackExo* body and the movement between these bodies is rotational ranging among -1.74533 to 0.349066 radians. For another words, this is the definition of the flexion/extension movement of the exoskeleton right hip.

```
<PinJoint name="Exo_hip_r">
    <!--Name of the parent body to which this joint connects its owner body.-->
    <parent_body>BackExo</parent_body>
    <!--Location of the joint in the parent body specified in the parent reference frame. Default is (0,0,0).-->
    <location_in_parent>0.07804111 -0.045425 0.175</location_in_parent>
    <!--Orientation of the joint in the parent body specified in the parent reference frame. Euler XYZ body-fixed rotation angles are used to express the orier
    <orientation_in_parent>0 0 0</orientation_in_parent>
    <!--Location of the joint in the child body specified in the child reference frame. For SIMM models, this vector is always the zero vector (i.e., the body
    <location>0.07804111 -0.045425 0.175</location>
    <!--Orientation of the joint in the owing body specified in the owning body reference frame.  Euler XYZ body-fixed rotation angles are used to express the
    <orientation>0 0 0</orientation>
    <!--Set holding the generalized coordinates (q's) that parmeterize this joint.-->
    <CoordinateSet>
        <objects>
            <Coordinate name="Exo_hip_r">
                <!--Coordinate can describe rotational, translational, or coupled motion. Defaults to rotational.-->
                <motion_type>rotational</motion_type>
                <!--The value of this coordinate before any value has been set. Rotational coordinate value is in radians and Translational in meters.-->
                <default_value>0</default_value>
                <!--The speed value of this coordinate before any value has been set. Rotational coordinate value is in rad/s and Translational in m/s.-->
                <default_speed_value>0</default_speed_value>
                <!--The minimum and maximum values that the coordinate can range between. Rotational coordinate range in radians and Translational in meters.--
                <range>-1.74533 0.349066</range>
                <!--Flag indicating whether or not the values of the coordinates should be limited to the range, above.-->
                <clamped>false</clamped>
                <!--Flag indicating whether or not the values of the coordinates should be constrained to the current (e.g. default) value, above.-->
                <locked>false</locked>
                <!--If specified, the coordinate can be prescribed by a function of time. It can be any OpenSim Function with valid second order derivatives.--
                <prescribed_function />
                <!--Flag indicating whether or not the values of the coordinates should be prescribed according to the function above. It is ignored if the no
                <prescribed>false</prescribed>
                <!--Flag identifies whether or not this coordinate can change freely when posing the model to satisfy kinematic constraints.  When true, the cc

            </Coordinate>
```

Figure 2.4 - Code segment where a Pin joint is defined. It includes definition of parent body, location, orientation, and coordinate.

The body definition finished with the GeometrySet where the CAD files were assigned to each body and after that the model was ready to be visualized in the GUI. Lastly, with the help of the GUI view, it was necessary to adjust the “location in parent parameter” in order to correct the relative positions of the bodies.

## 2.3 Model Validation

Validation verifies the accuracy of a model, i.e., it is the process that confirms that the modelling is correct and the simulation represents well the system behavior.

The general workflow used in OpenSim projects is presented in Figure 2.5. First, the model is scaled in order to be in agreement with experimental data. Inverse Kinematics and Inverse Dynamics represent the tests performed to validate the model in terms of its kinematics and dynamics.


Figure 2.5 - Typical workflow used in OpenSim: First, the model is scaled, afterwards Inverse Kinematics simulation followed by Inverse Dynamics simulations are performed.

OpenSim has a Scale tool that allows the user to scale a model so it can fit with experimental data aiming to get good results in following tools as Inverse Kinematics. In the *Exoskeleton* model, this step was done during the modelling process using SolidWorks.

H1 is equipped with several sensors which provide very meaningful information. In particular, the exoskeleton has a position sensor in each joint measuring the joint angle and a torque sensor which measures the torque in each joint. Therefore, to validate the model two tests were made using Inverse Dynamics tool.

The common process would be to proceed with Inverse Kinematics after the scaling, a tool that estimates joint angles based on measured trajectories of skin-mounted markers. After that, the calculated joint angles would be used as input to the Inverse Dynamics tool. However, since H1 sensors determine joint angles instead of Cartesian coordinates which would be the right input for Inverse Kinematics, this step was skipped and the test used to validate the model was Inverse Dynamics.

The model validation started with a static validation. Besides reading the joint angles, H1 can also follow instructions and move its joints to achieve a requested position. Thus, with H1 fixed to its support, several positions were tested with the goal of reading the joint moments. The positions tested in the trial are presented in Table 2.2.

Table 2.2 - Definition of the positions tested in H1. All values are in degrees.

| Positions | Right Hip | Right Knee | Right Ankle | Left Hip | Left Knee | Left Ankle |
|---|---|---|---|---|---|---|
| **P0** | 0 | 0 | 0 | 0 | 0 | 0 |
| **P1** | 50 | 0 | 0 | 50 | 0 | 0 |
| **P2** | 50 | -50 | 0 | 50 | -50 | 0 |
| **P3** | 50 | -50 | -15 | 50 | -50 | -15 |
| **P4** | 80 | 80 | 0 | 80 | 80 | 0 |
| **P5** | 78 | 10 | 0 | 20 | 20 | 80 |

After collecting the experimental data, it was prepared as input for Inverse Dynamics. This tool receives motion files (.mot) as input, and this file type was created by SIMM developers. Motion files include two parts: header and data. Header contains the generic information of the data, as how many rows and columns are in the file, and data units. When the header ends, the data begins. Usually, the first column is time, and the next columns contain values that vary with time such as generalized coordinates, marker coordinates, ground reaction forces and moments, centers of pressure, muscle activations, or muscle lengths. Figure 2.6 presents an example of a motion file.

```
Coordinates
version=1
nRows=10
nColumns=7
inDegrees=yes

"Units are S.I. units (second, meters, Newtons, ...)"
Angles are in degrees.

endheader
time	Exo_hip_r	Exo_knee_r	Exo_ankle_r	Exo_hip_l	Exo_knee_l	Exo_ankle_l
0	0	0	0	0	0	0
0.01	0	0	0	0	0	0
0.02	0	0	0	0	0	0
0.03	0	0	0	0	0	0
0.04	0	0	0	0	0	0
0.05	0	0	0	0	0	0
0.06	0	0	0	0	0	0
0.07	0	0	0	0	0	0
0.08	0	0	0	0	0	0
```

Figure 2.6 - Motion file of Position 0. The file contains the joint angle coordinates for each joint of the *Exoskeleton* model of the Position 0. It is divided in header and data.

After the data was prepared, the model was opened in OpenSim GUI and then, the Inverse Dynamics tool was initiated. This tool computes the net forces and moments at each joint underlying a particular movement [47].

Inverse Dynamics is based on the classical equation of motion, that can be written, in the inverse dynamics sense [48], as follow:

$$M(q)\ddot{q} + C(q,\dot{q}) + G(q) = \tau$$

Equation 1

Being N the number of DOF, $q, \dot{q}, \ddot{q} \in R^N$ are vectors of generalized positions, velocities and accelarations, respectively; $\mathrm{M(q)} \in R^{NxN}$ is the system mass matrix; $\mathrm{C}(\mathrm{q}, \dot{q}) \in R^N$ is the vector of Coriolis and centrifugal forces; $G(\mathrm{q}) \in R^N$ is the vector of gravitational forces and $\tau \in \mathbb{R}^N$ is the vector of generalized forces that must be determined. All the terms of the left side of the equation are known, since the motion of the model is completely defined by the position, orientation, velocity, and acceleration of the center of mass of the segments. Thus, inverse dynamics solves the equation in order to determine the right-side term, $\tau$, the net forces and torques for each joint.

H1 is capable of following a given trajectory, so, considering this, another trial was created. This trial consisted in giving the exoskeleton a recorded trajectory of a healthy subject, named *gait demo* and as in static validation it recorded data of joint angles and joint moments. An acquisition of 1000Hz during 40 seconds was done. The following steps were similar to static validation, the data were processed and corresponding motion files were created.

Figure 2.7 presents a screen capture of the OpenSim environment where the exoskeleton model is represented, and in the first plan shows Inverse Dynamics Tool with all the parameters needed to *Run* (execute the tool) defined, the input file, the filter frequency, time range and the output directory.

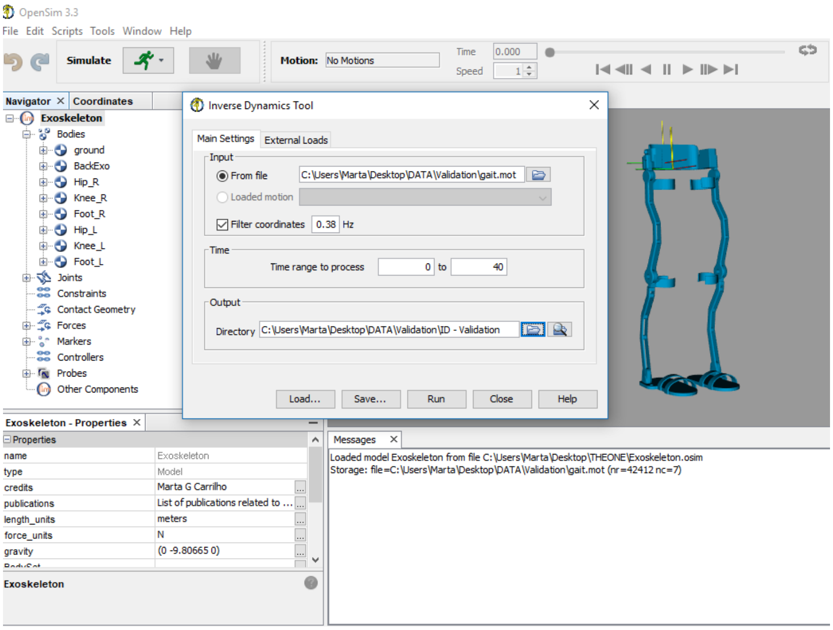



Figure 2.7 - OpenSim environment including the Inverse Dynamics tool. In the background, the OpenSim environment where is possible to see the model and the list of its components.

## 2.4 Exoskeleton Integration

The last part of this project was the integration of the exoskeleton model and a musculoskeletal model in a new model, named *ExoBody* model, with the goal of studying the interaction between both of them.

The musculoskeletal model chosen to be part of the new model was the *3DGaitModel2392* [49]. This model is composed of head, torso and lower limbs, having 92 musculotendon actuators and 23 DOFs. The DOFs included in the model consist in: pelvis position (3 DOFs), pelvis orientation (3 DOFs), lumbar joint (3 DOFs), and another DOF for each leg, hip flexion–extension, abduction–adduction, and internal–external rotation, knee flexion–extension, ankle plantarflexion–dorsiflexion, inversion-eversion and toes flexion-extension.

The *ExoBody* model contains all the bodies of the exoskeleton model and of the *3DGaitModel2392*, keeping its corresponding joints. Connecting both systems implies that puts limits to each other, when one moves, the other must follow and vice versa. Due to this, it is necessary to include constraints in this model.

OpenSim supports 3 types of constraints:

- Coordinate Coupler constraint - relates the generalized coordinate of a given joint (the dependent coordinate) to any other coordinate in the model (independent coordinate);
- Point Constraint – fixes a point defined with respect to the two bodies (no relative translations);
- Weld Constraints - fixes the relative location and orientation of the two bodies (no translations or rotations).

Three weld constraints were applied in the model: one at the level of the waist between the pelvis bone and the *BackExo* body, and two at foot level. The adjustment of relative positions of exoskeleton and musculoskeletal system was made by try-error.

When the *ExoBody* model was ready, a gait study was performed using data from the OpenSim database [50]. The main goal of this study was to evaluate the *ExoBody* model in simulation, verifying if there were errors or anything unusual. But, it is important to refer that this was not a validation process.

Hence, two different data sets were used, both containing positions of skin-mounted markers, one referent to a stroke patient while referred to healthy individual. The conditions of the experimental trials in which these data were recorded were not known with exception of the markers initial position. The data sets were already prepared for OpenSim tools. Thus, the study consisted in comparing the gait of a stroke patient with a healthy gait, through Inverse Kinematics and Inverse Dynamics.

Inverse Kinematics tool estimates joint angles based on measured trajectories of skin-mounted markers. For each frame, it is solved a least-squares problem, in order to minimize the weighted error for each coordinate [51]:

$$Squared\ Error = \left[\sum_{i\ \in markers} w_i \left\| x_i^{exp} - x_i(q) \right\|^2 + \sum_{j\ \in unprescribed\ coords} w_j \left(q_j^{exp} - q_j\right)^2\right]$$

Equation 2

Where, q is the vector of the generalized coordinates being solved for, $x_i^{exp}$ is the experimental position of the $i^{th}$ marker, $x_i(q)$ is the position of the corresponding marker on the model (which depends on the coordinate values), $q_j^{exp}$ is the experimental value for coordinate j. $w_i$ and $w_j$ are the corresponding weights of the markers and the joints.

The kinematics tool receives as input a .trc file containing the experimental coordinates of the markers, and its output is a .mot file with the joint angles. The output file of inverse kinematics is usually used as input in inverse dynamics.

# 3. RESULTS

In this chapter, the results are presented. The first section presents the results of Inverse Dynamics and consequent validation of the model. The next section presents the Inverse Kinematics and Inverse Dynamics results in the context of gait simulation analysis performed with the *ExoBody* model.

The view of the model constructed is presented in Figure 3.1 in different view perspectives. In Figure 3.2 it is possible to see a summary of the model constitution, the topology view presents all the bodies, in boxes, that compose the model and the joints between them represented by the arrows connecting the boxes. The *Exoskeleton* can perform simulations within few minutes in a typical desktop computer. Moreover, it is possible to adjust parameters as the height, the weight, in order to personalize the model, as well as to add actuators or external forces.

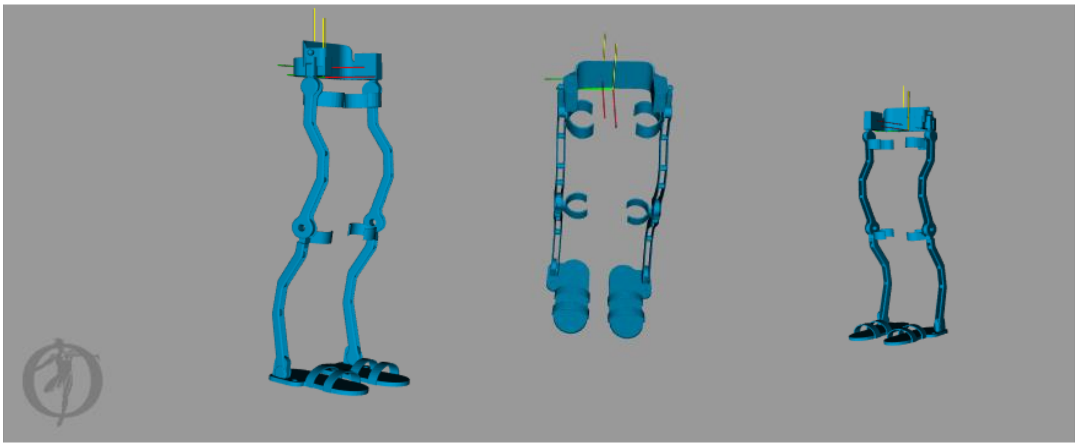


Figure 3.1 - Exoskeleton model in different perspectives. Screen capture of OpenSim environment.

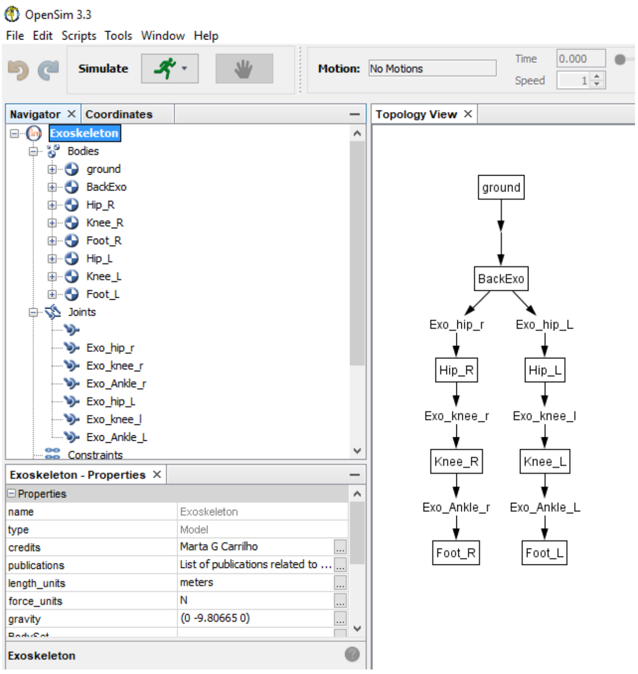



Figure 3.2 - Model resume. OpenSim environment: In Navigator the components of the model are presented and the topology view represents all these elements and the connections between them.

## 3.1 Validation

When the modelling process was finished it was necessary to verify the accuracy of the model, i.e., confirming that the system is represented well and it acts in consistency with the system behavior. This process is known as validation process.

The model validation was made in two different ways: static validation and motion validation. For the first process, several trials were performed, with the exoskeleton in different positions but without moving, while the second concerned only one trial where the exoskeleton was moving. The validation goal is equal for both static and validation process, consisting in comparing experimental data with the results of Inverse Dynamics tool expecting them to be similar in order to validate the model. Even though the methods for both validations were similar, the results will be presented in two separated subsections of this chapter.

### 3.1.1 Static Validation

The static validation considers the trials performed in H1 where the positions defined in Table 2.2 were tried on the wearable device. The result of these trials was the experimental joint moments measured with the H1 sensors that are presented in the Table 3.1.

Table 3.1- Joint moments reported by H1 sensors. All values are in Nm.

| Positions | Right Hip | Right Knee | Right Ankle | Left Hip | Left Knee | Left Ankle |
|---|---|---|---|---|---|---|
| **P0** | 3.35 | -0.9 | 0.9 | -0.25 | -0.85 | 2.7 |
| **P1** | -6.9 | -8.8 | 1.2 | 5 | 5.6 | 3.3 |
| **P2** | -2.8 | -0.05 | 1.1 | 4.7 | -1.2 | 3.3 |
| **P3** | -2.6 | 0.1 | 1.2 | -2 | 4.7 | 2.4 |
| **P4** | -5.1 | -0.3 | 1.2 | 4.7 | -2.2 | 2.8 |
| **P5** | -10 | -11 | 1 | -2.3 | -8.6 | 2 |

In the model, the joint moments were calculated with Inverse Dynamics as explained in section 2.3. Using the positions of table 2.2 as input, the results obtained are presented in Table 3.2.

Table 3.2 - Joint moments calculated by Inverse Dynamics tool. All values are in Nm.

| Positions | Right Hip | Right Knee | Right Ankle | Left Hip | Left Knee | Left Ankle |
|---|---|---|---|---|---|---|
| **P0** | -1.89836 | -1.46949 | 0.33759 | -1.21586 | -1.51204 | 0.34794 |
| **P1** | 14.69594 | 4.63047 | 0.46897 | 15.15305 | 4.63398 | 0.47056 |
| **P2** | 8.59599 | -1.46949 | 0.33759 | 9.00702 | -1.51204 | 0.34794 |
| **P3** | 8.49935 | -1.56612 | 0.24095 | 8.91175 | -1.60732 | 0.25267 |
| **P4** | 11.75037 | -1.46949 | 0.33759 | 11.81769 | -1.51204 | 0.34794 |
| **P5** | 19.31253 | 6.19728 | 0.43144 | -2.20376 | -7.09359 | -0.10516 |

The comparison of the experimental joint moments with the ones calculated with Inverse Dynamics is represented in Figure 3.3. Each graphic corresponds to one position, and represents the joint moment, in Nm, for each joint which are represented by numbers: (1) Right Hip; (2) – Right Knee; (3) – Right Ankle; (4) – Left Hip; (5) – Left Knee; (6) – Left Ankle. The blue line is regarding to experimental data obtained from the exoskeleton sensors, while the orange line corresponds to the data obtained from OpenSim tool.

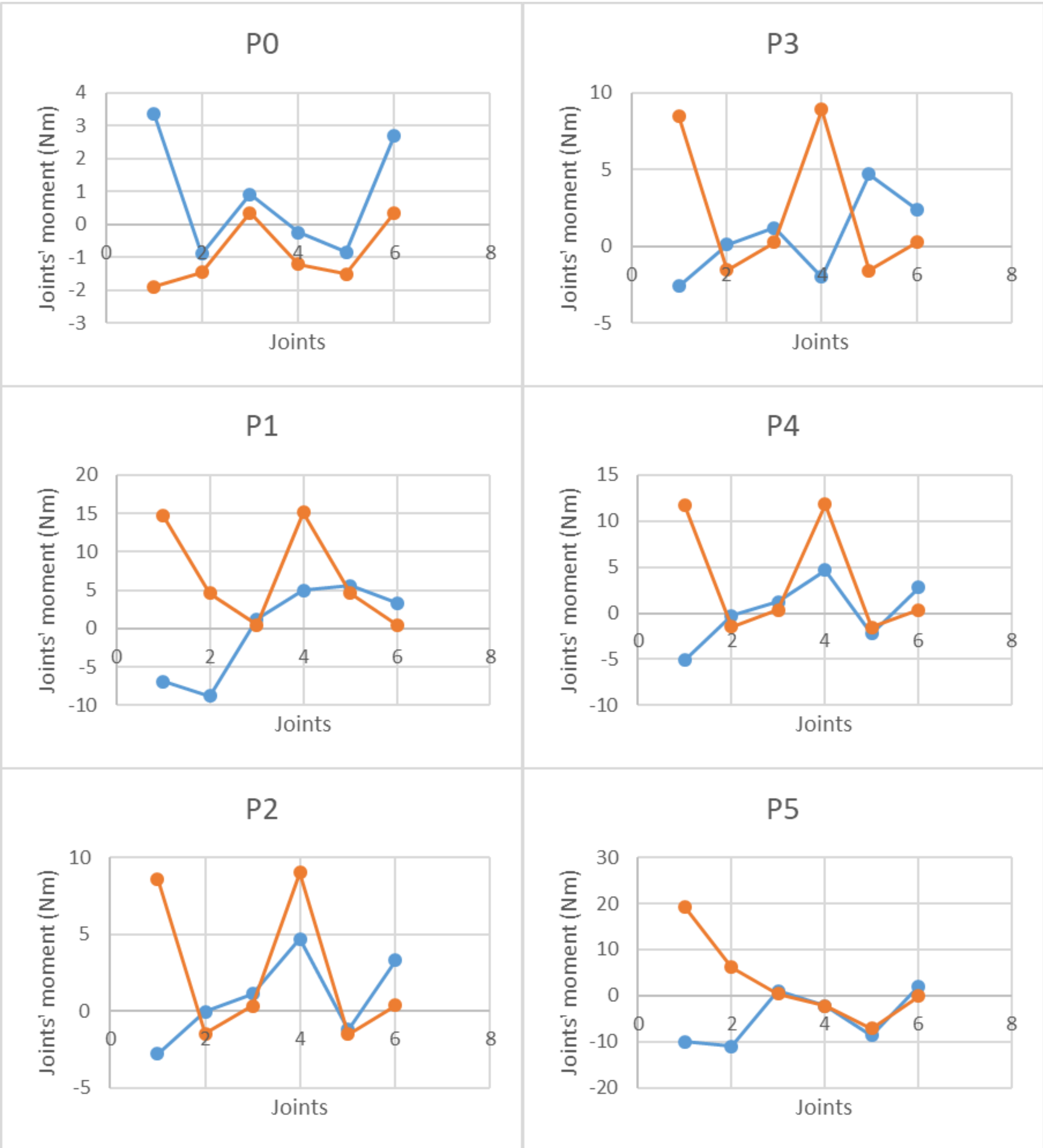



Figure 3.3 Joint moments comparison: The graphics represent the joint moments for each joint, corresponding each graphic to the positions of table 2.2. The blue line represents the experimental data from table 3.1 whilst the orange line corresponds to the joint moments calculated with Inverse Dynamics, table 3.2.

### 3.1.2 Motion Validation

Similarly to the static validation, the procedure of motion validation consisted in comparing the joint moments recorded by the H1 sensors with those calculated with Inverse Dynamics.

The gait of an unknown subject was recorded and the trajectory was used as control-trajectory. Fixed to its support, the exoskeleton played this trajectory in order to record the values of the joint moment. As described in the previous chapter, an acquisition of 40s was made. The joint angles used as input of Inverse Dynamics tool were not filtered before they were used in OpenSim, however, they were filtered during Inverse Dynamics simulation. A low pass filter of 0.5Hz was used, even though it was not the best frequency for all the joints, it was the one which adapted best for all. The Figure 3.4 presents the comparison of experimental joint moments with those calculated with Inverse Dynamics tool.

Although the trial took 40 seconds, the graphics only represent 20 seconds for two reasons: (1) it is representative, the following 20 seconds represented another four cycles of gait. (2) the graphic is clearer, allowing for a better comprehension.

Each plot refers to a specific joint, and it represents the join moment, in Newton Meter (Nm) in function of time, in seconds. The red line, named *OpenSim*, corresponds to the joint moments obtained through Inverse Dynamics using as input the joint angles recorded with the exoskeleton. On the other hand, the blue line, named *Experimental*, refers to the experimental joint moments directly recorded by the exoskeleton sensors.

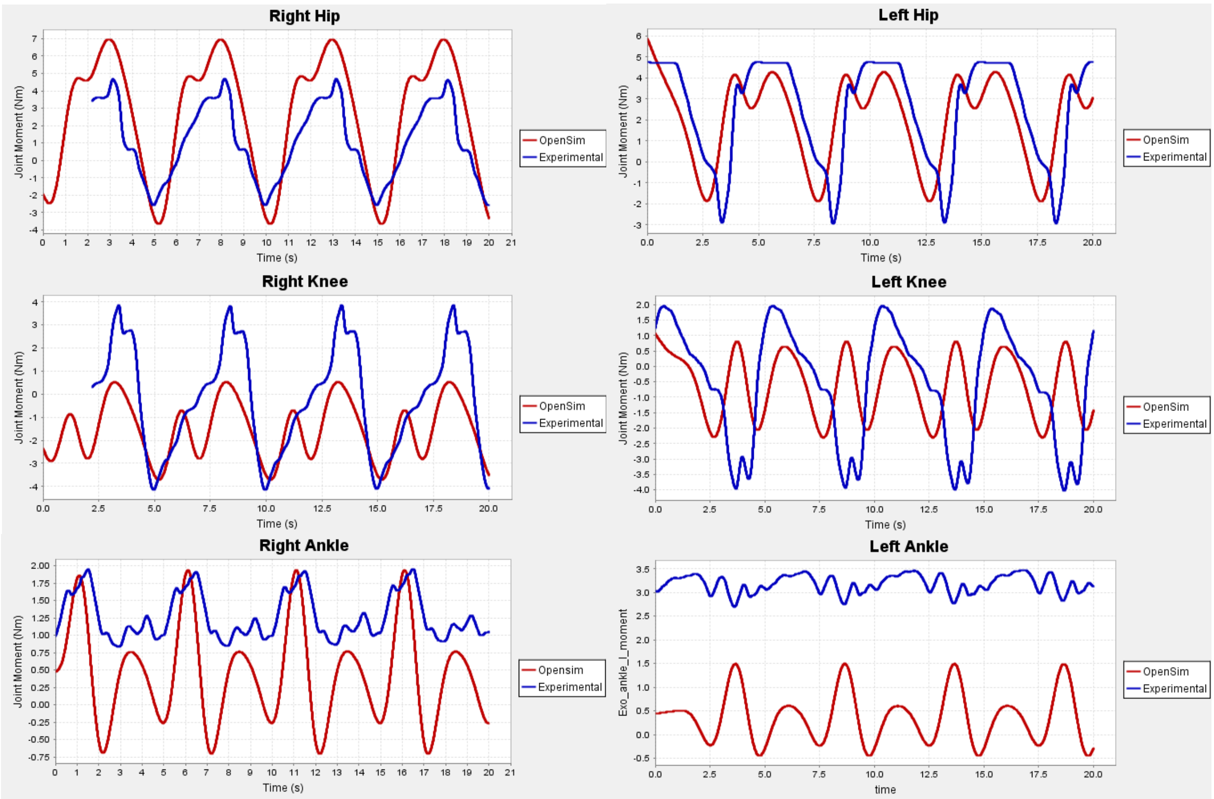



Figure 3.4 - Motion validation: Joint moments comparison. Each graphic refers to a joint, representing the joint moment in Nm, in function of time in seconds. The red line refers to simulated data while the blue line to experimental data.

## 3.2 GAIT ANALYSIS

A new model, the *ExoBody*, was created including the exoskeleton model and an existing musculoskeletal model. *ExoBody* is presented in Figure 3.5.

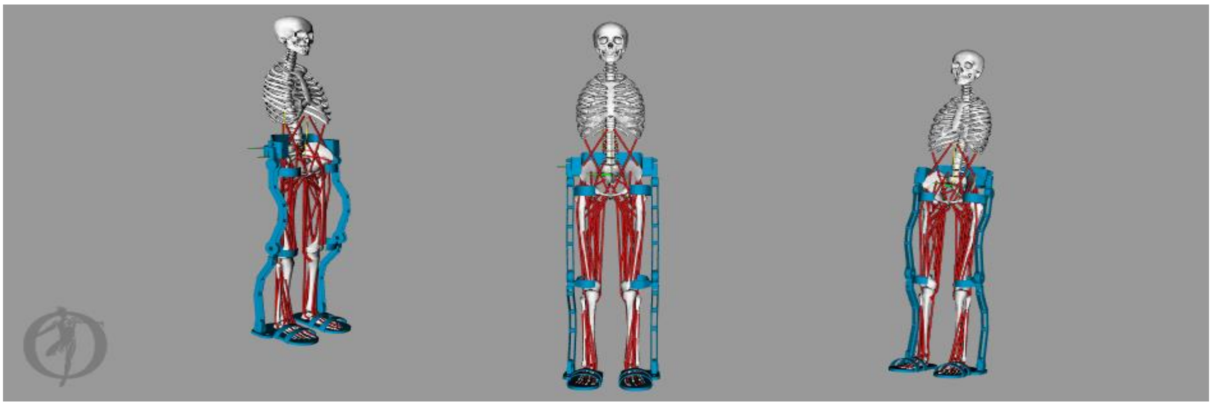


Figure 3.5 - ExoBody Model: the model includes the exoskeleton model and the 3DGait2392 model.

This model enables the possibility of studying the interaction between robotic device and patient. To understand that, a gait analysis was done. The data sets available had a short time duration. The following graphics presents the kinematics and dynamics gait study of two subjects: a healthy one and a stroke patient, both studied with and without exoskeleton.

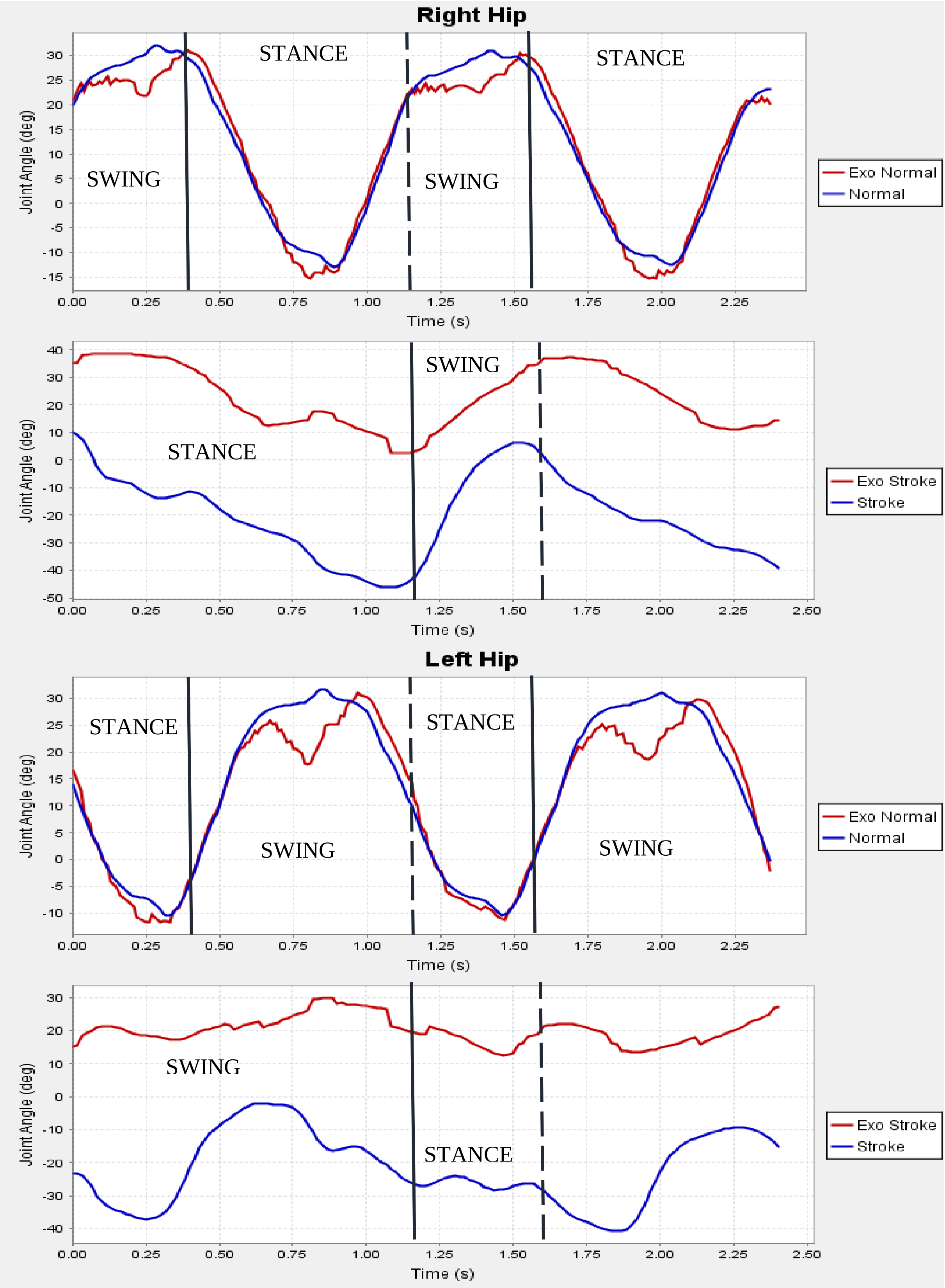



Figure 3.6 - Hip Joint: results of the kinematics analysis. The two-upper plots refer to the right joint and the two-lowest to the left side. The first and third graphic represent a normal pattern whereas the second and the fourth correspond to a hemiplegic gait pattern.

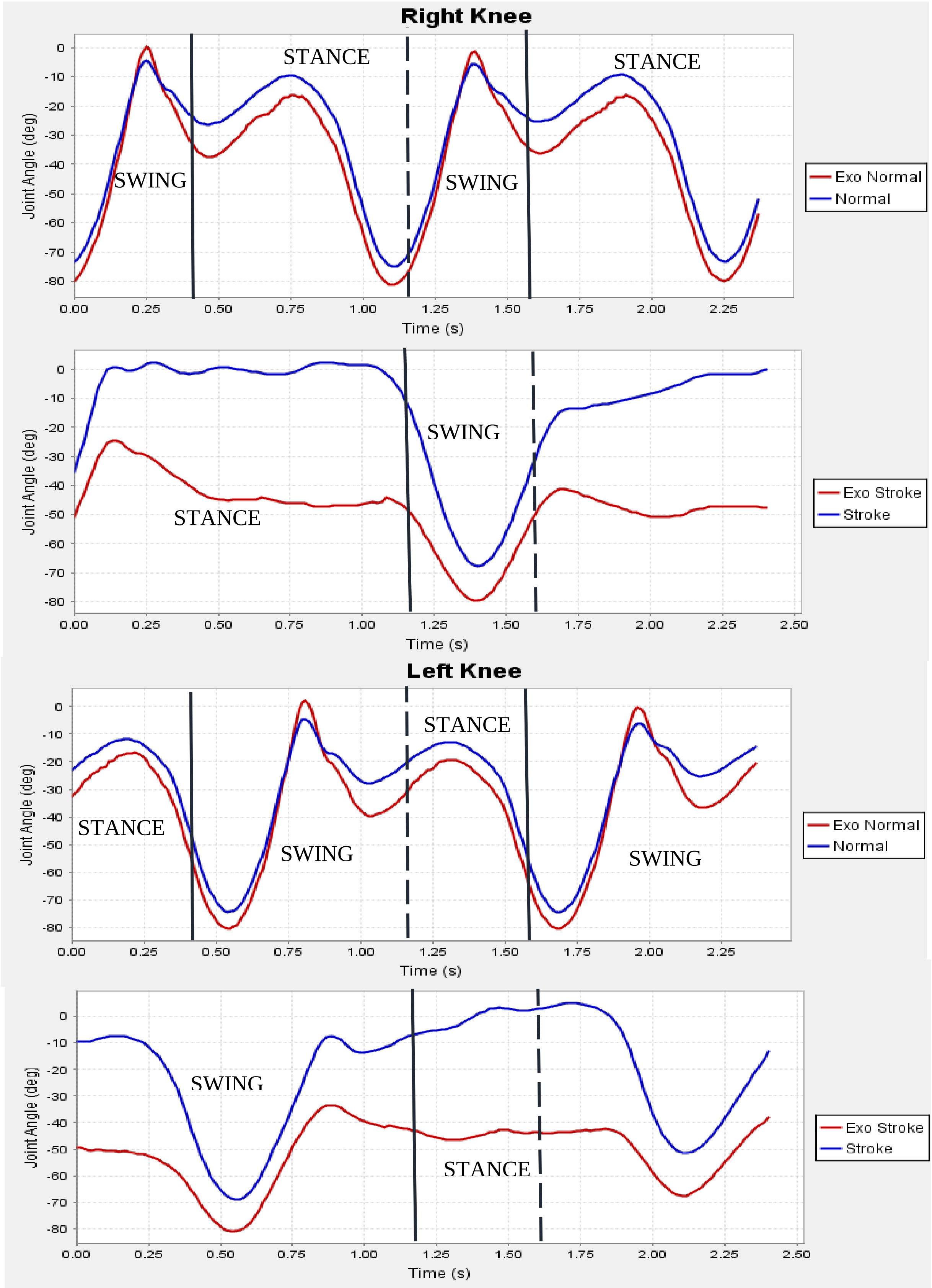



Figure 3.7 - Knee Joint: results of the kinematics analysis. The two-upper plots refer to the right joint and the two-lowest to the left side. The first and third graphic represent a normal pattern whereas the second and the fourth correspond to a hemiplegic gait pattern.

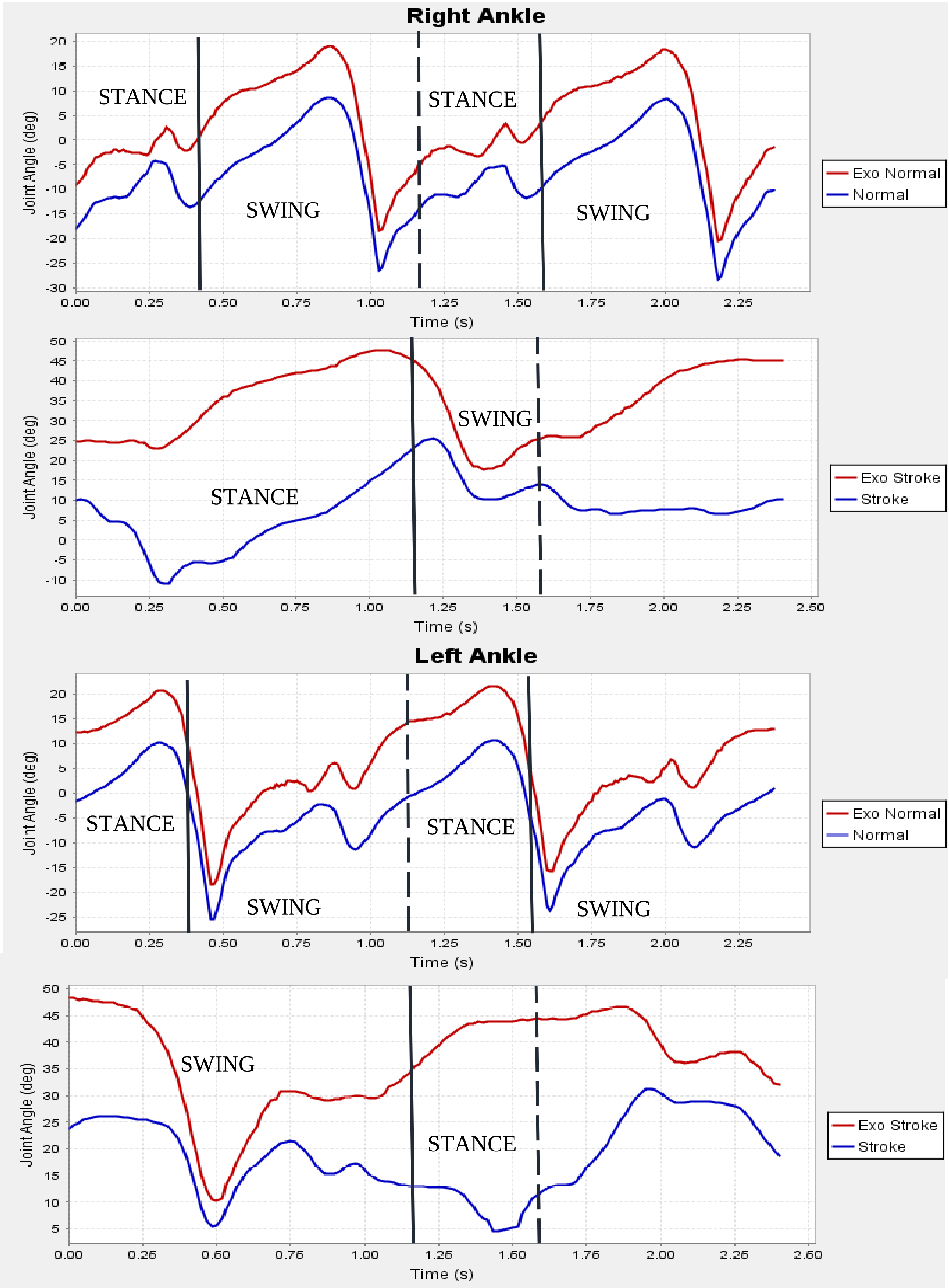



Figure 3.8 - Ankle Joint: results of the kinematics analysis. The two-upper plots refer to the right joint and the two-lowest to the left side. The first and third graphic represent a normal pattern whereas the second and the fourth correspond to a hemiplegic gait pattern.

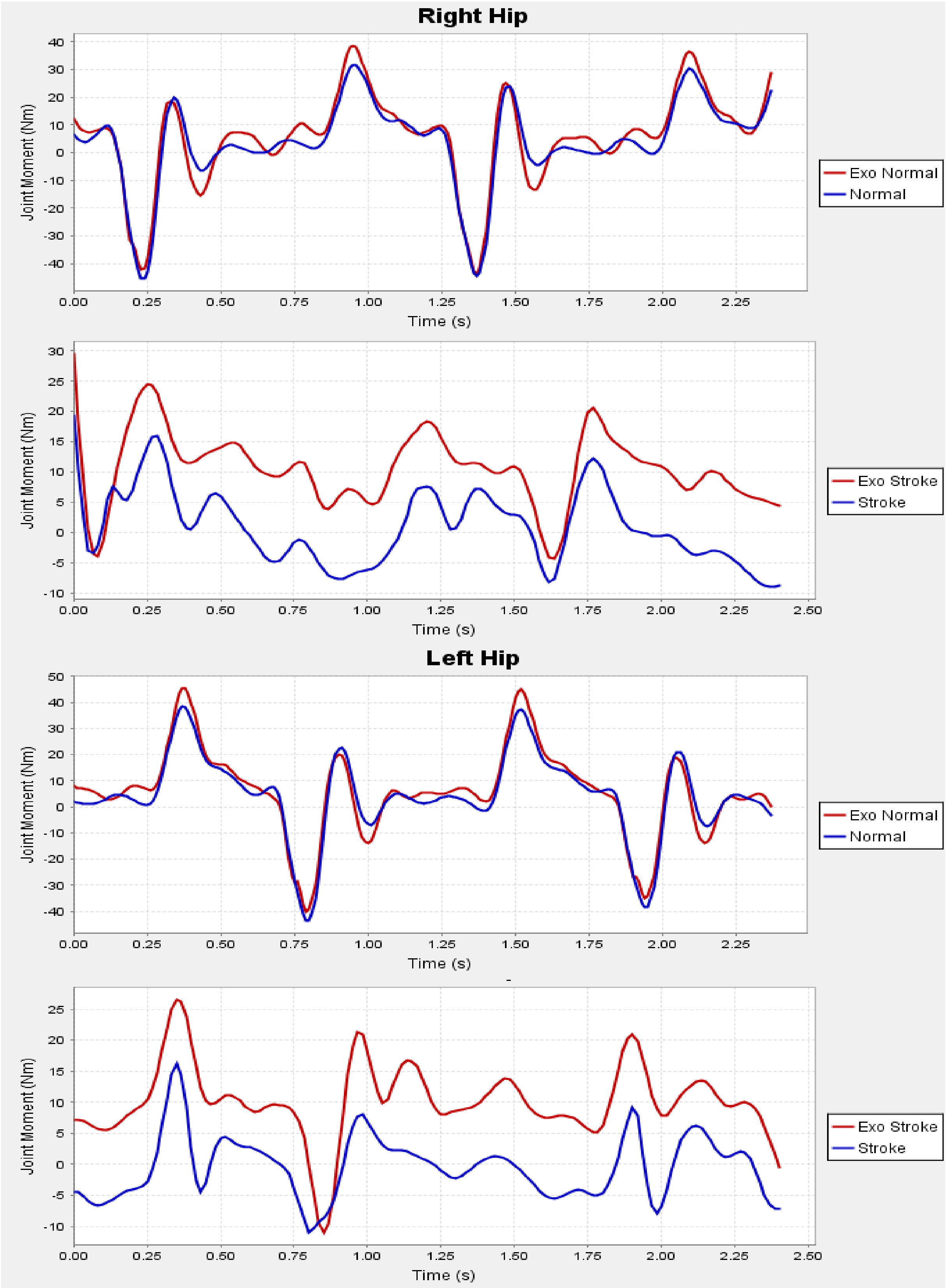



Figure 3.9 - Hip Joint: All graphics present Joint Moment, in Nm, versus time, in s. The two-upper graphics refer to the right joint and the two-lowest to the left side. First and third plots regarding to Normal gait pattern while the second and the fourth represent Hemiplegic pattern.

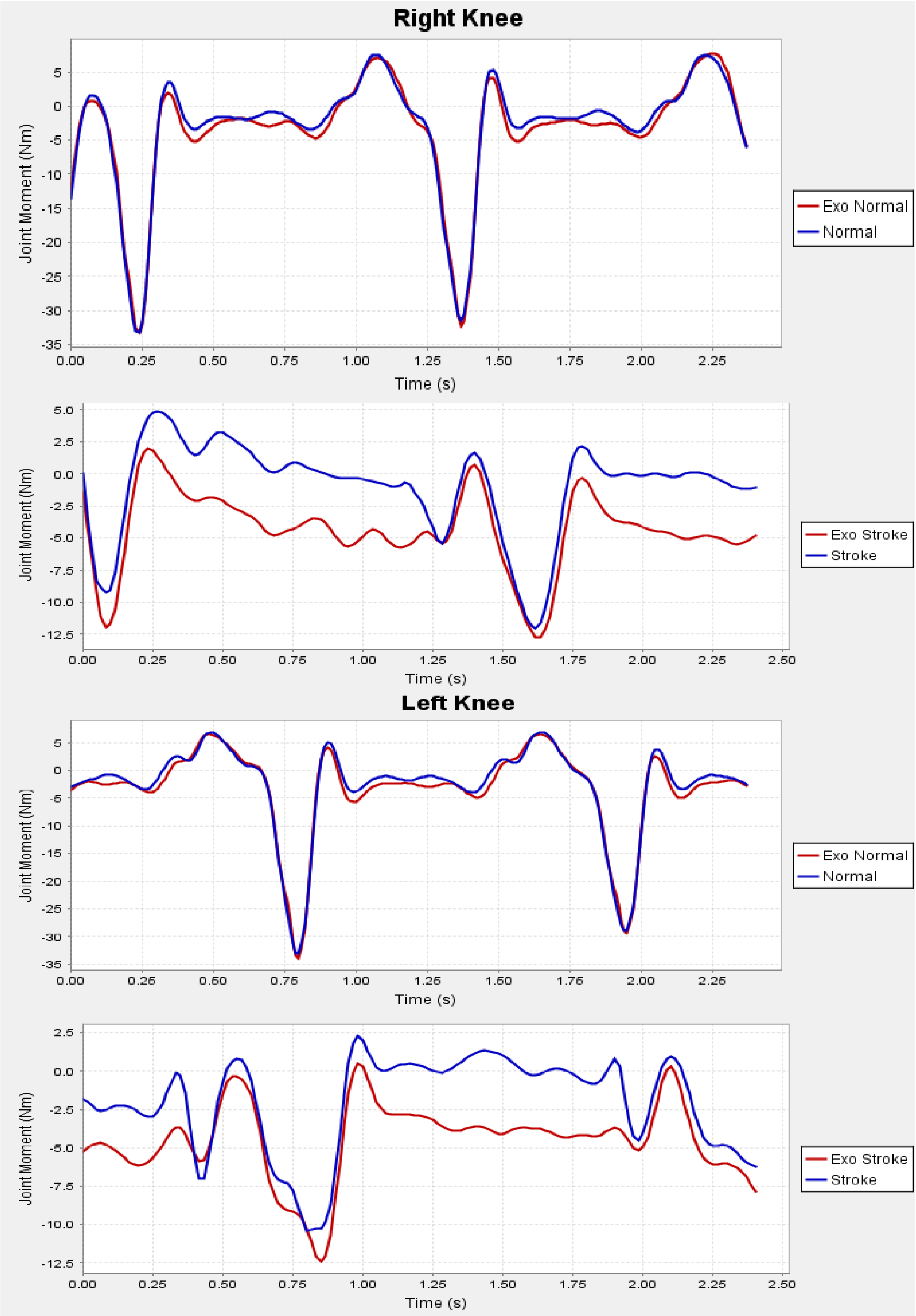



Figure 3.10 - Knee Joint: All graphics present Joint Moment, in  Nm, versus time, in s. The two-upper graphics refer to the right joint and the two-lowest to the left side. First and third plots regarding to Normal gait pattern while the second and the fourth represent Hemiplegic pattern.

Each figure contains the graphics of one joint, the two-upper are regarding to the right joint and the two-lowest refer to the left joint. The *Exo Normal* and *Exo Stroke* curves are regarding to simulations performed with the *ExoBody* model while *Normal* and *Stroke* curves concern the *3DGait2392* model simulations. Figure 3.6, Figure 3.7 and, Figure 3.8 present the Joint Angles in degrees (deg) versus Time, in seconds (s) calculated with Inverse Kinematics. Figure 3.9, Figure 3.10 and, Figure 3.11 represent the Joint Moment, in Newton meter, in function of time, in seconds. Although the stroke patient data set had a duration of 5 seconds, about 2 seconds more than the healthy subject, only 2.4 seconds are presented in graphic so it is possible to compare in the same time reference.

# 4. DISCUSSION

Chapter 4 focusses on the analysis and discussion of the results presented. First, the results of the modelling and validation processes are discussed. More specifically, the experimental data acquired in trials with H1 is compared with the results of Inverse Dynamics simulation that used the joint angles as input. The next section contains the gait analysis of two subjects, comparing a healthy subject with a stroke patient.

## 4.1 MODELLING AND VALIDATION PROCESSES

A representative model of the H1 exoskeleton that can be used to simulate this device's behavior in different conditions was created. This model is capable of performing kinematic and dynamic simulations within few minutes in a typical desktop computer. To prove its accuracy and reliability a validation process was performed and the results are discussed in the following paragraphs.

Regarding to Static Validation, the comparison between experimental and simulated data is presented in Figure 3.3, where the blue line regards to experimental data while the orange line refers to simulated data. In all the graphics presented, the joint moments are similar for almost every joint with exception of joints 1 and 4 which correspond to the right and left hip joints, respectively. These joints reveal a significant difference between the joint moment recorded by the exoskeleton sensor and the joint moment calculated in OpenSim. It is believed that this difference is due to hardware problems, more specifically problems in some potentiometers which might not have been well calibrated or had problems in the connection. Furthermore, as explained the trials were performed with the exoskeleton fixed in its support and this could also influence the results. As H1 was being held by its waist, the hip joints were the most susceptible to suffer alterations in joint moment since they were supporting more weight.

The results of the motion trial are presented in Figure 3.4. In general, comparing the experimental data (blue curves) with the simulated data (red line), the curves have a similar shape, however, they show different range values. It is important to refer that in the right hip and knee plots, the experimental data was normalized in time reference because the sensors of the mentioned joints did not start recording the joint moments right at the beginning of the movement.

Considering the hip joint, both graphics present a good agreement between experimental and simulation data. Regarding the knee joints, there are few differences, especially in the left knee. Although the curves are well synchronized in time reference, i.e., the GC duration is the same, the curves shape is not so well synchronized as in hip joints. The most probable cause for this is the low pass filter, with a cut off frequency of 0.5 Hz, used in the Inverse Dynamics tool. Comparing both sides it is possible to see that the range of values for experimental data is analogous proving that the gait is symmetric, but this does not happen with the simulated joint moments.

The ankle joint results are the ones which present less resemblances between experimental and simulated data. Regarding the right ankle joint, the curves are coordinated in time. The maximum peak of the joint moment is nearly coincident, though the curve obtained from OpenSim has an amplitude larger than the experimental curve. In the left joint is harder to find similarities between the curves. This is the only plot of Figure 3.4 where the curves don't even cross each other. Furthermore, relating the right ankle joint with the left one, the experimental data are not coincident in the range of values, i.e., the gait performed is not symmetric. However, comparing the simulated data, the curves shape is similar and the amplitude is equal. Even though some differences exist between the experimental joint moment curves, it is important to note that, these curves have several peaks within short time intervals and considering the low pass filter applied, it is normal that the simulated curves have less peaks.

## 4.2 Gait Analysis

The *ExoBody* model, which includes the *Exoskeleton* model and the *3DGait2392* model, was designed to enable the study of the interaction between both systems: human and robot. The model showed the capability of performing kinematic and dynamic simulations in few minutes in a typical desktop computer, however, this model was not validated because the validation process would involve a complex study of ground reaction forces which was not possible to realize during this project. Even though a few simulations were performed with data from the OpenSim database, allowing to do a gait analysis comparing two subjects: a healthy one and a stroke patient. The kinetics and dynamics results are presented in Figure 3.6 to 3.9 and 3.10 to 3.11, respectively.

To begin with the kinematic analysis, comparing Figure 3.6, Figure 3.7, and Figure 3.8 it is possible to see that the curves regarding the normal gait pattern are similar in shape and in range of values for both models. However, regarding the hemiplegic pattern the shape is also similar but there are differences in joint angles range of values. As expected, the GC duration, whose end is defined by the dashed lines, is superior in the hemiplegic gait pattern.

The results of the hip joint for the healthy subject are in agreement with the reference presented in Figure 1.4. The results concerning the hemiplegic gait pattern have a reduced amplitude, also, it was expected a decrease of hip extension but it is not verified. Even though the joint angles are out of range and not close to the reference, there is a notable difference when the simulation is performed with the *ExoBody* model.

Figure 3.7 contains the knee joint angles plots that have a common characteristic: in all graphics the maximum angle is near 0 whereas in the reference the knee joint angle curve is always above 0º. Considering the absolute values of the knee joint angles for the normal gait pattern, the curves would have similar shape and range of values, with the only exception of the maximum peak of the knee extension. Regarding the hemiparetic subject, also considering absolute values, there are some characteristics of the pattern possible to identify as the reduced knee flexion during early-stance phase, although not followed by knee-hyperextension as mentioned in [7]; a knee flexion peak of approximately 65º in swing phase; also during swing phase, a general decrease in knee flexion is observed. Despite of what happens with the hip and ankle joints, in the knee joint the blue curve is above the red curve, but considering absolute values the knee joint would maintain the pattern highlighting the calculated joint angles with the *ExoBody* model are larger than the ones calculated with the *3DGait2392*. Furthermore, for the knee joint the simulation performed with *ExoBody* model with the stroke patient data – Exo Stroke curve – is closer to the simulation of the same model with the healthy subject data.

To finalize the kinematics analysis, Figure 3.8 presents the ankle joint angles. The normal gait pattern results are in agreement with the reference. Taking into consideration the hemiplegic pattern, a decrease in dorsiflexion was expected but instead there is an increase. The results are not in agreement with the reference.

Regarding the study of dynamics, the joint moments are presented in Figure 3.9, Figure 3.10 and Figure 3.11 and the reference is presented in Figure 4.1. As the hip joint angles of the healthy subject, the hip joint moments are also in agreement with the reference. The right and left hip moments are complemental, strengthening the gait symmetry. There is not much difference between the simulation with the *ExoBody* model or with the *3DGait2392*, which might prove the accuracy of the *ExoBody* model. Making allowance to the hemiplegic pattern the joint moments are lower in comparison with the healthy subject which is common for stroke patients. The hip moments calculated within *ExoBody* model are larger than the ones calculated with the *3DGait2392* for the hemiplegic pattern. Although the knee joint angles were out of range, with the knee joint moments the situation is not equal. Regarding the healthy subject, the curves shape is alike the reference, however the minimum peaks are lower than the reference. In the same way, as with the hip joint the joint moments of the stroke patient are reduced comparing with the healthy control but in the knee joint the difference between the *ExoBody* simulation and the *3DGait2392* is smaller. Moreover, the range of values has different maximum peak for each side, confirming the right limb as paretic limb. The ankle joint moment results, for the healthy subject are not in agreement with the reference: the plots present more peaks than the reference and the range of values is above the expected. The hemiplegic pattern results also present several peaks, especially the simulations done with the *ExoBody* model. Nonetheless, the *Exo Stroke* curve has a similar shape to the *Exo Normal*.

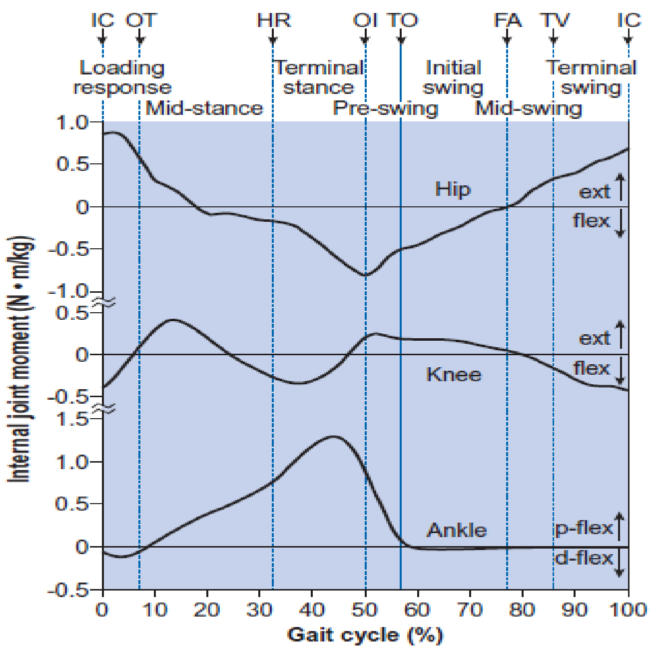



Figure 4.1 - Joint Moment reference. Image from [41]

In brief, not all the results are in accordance with the references, but most of the results are congruent with each other, showing gait symmetry as expected for the healthy subject or the gait cycle duration considering that in the hemiplegic pattern the GC has superior duration than in the normal gait pattern. The biggest limitation of the gait study was the lack of information about the subjects studied as well as the data processing done before their publication in the OpenSim database [50].

# 5. CONCLUSION AND FUTURE WORK

The main goal of this work was to develop an OpenSim model of an exoskeleton, more specifically, the H1 exoskeleton. The work started with an extensive study of the pathologies that might affect the human gait, as well as the state of the art of the available solutions, both presented in Introduction. From this study, it was concluded that until the present moment it is not available an open source model of an exoskeleton allowing a personalized gait analysis and treatment planning. Therefore, responding to this need, the *Exoskeleton* model was created and validated. Both static and motion validation presented an agreement between experimental and simulated data, confirming the model validation. Although in static validation, two joints exposed a significant difference between experimental and simulated data for two joints, it is believed that it is due to hardware problems, or the conditions in which the trials were performed and not related with the model itself. Thus, the main goal of this task was achieved, the H1 exoskeleton has an OpenSim model that allows the study of the robotic device behavior. This is the greatest advantage of this model, i.e., through simulation it is possible to improve the device performance, studying which variables should be improved or which parameters should be adjusted to optimize control strategies, or even to develop new control strategies. Nonetheless, the model also has liabilities as its design. The CAD model used was adapted in order to be very similar to H1, however it is not completely the same. Both static and motion validation showed some differences between the joint moment calculated with OpenSim and the experimental joint moment, these differences can be caused by scale difference.

After having the model validated, the next step was to integrate it in another model with a musculoskeletal model so the interaction between them can be studied. This step was achieved with success, the *ExoBody* model performed kinematics and dynamics simulations without problems. Considering that the *ExoBody* model includes two models, the *Exoskeleton* model and the *3DGait2392* model, both validated, theoretically, this model is also valid. However, more experiments should be done in the future, which are outside the scope of this project.

The gait study performed with the *ExoBody* model put in evidence the differences between a normal gait pattern and a hemiplegic gait pattern. Furthermore, the gait study exposed differences between the gait simulated with and without the exoskeleton. In general, the simulation of a stroke patient with the exoskeleton was closer to the healthy subject when simulated with the exoskeleton. Consequently, it is believed that this model can be very useful in robotics rehabilitation, since it enables a personalized gait analysis and the study of the interaction between patient and device. This can be a powerful tool in therapeutics plan and evolution analysis, allowing to adapt the treatment to each patient according to its disorder and or characteristics.

It is important to refer the limitations of the work developed. First, regarding the modelling process, several problems occurred during the model building, mainly due to the relative position of the model components which took more time than expected to solve. Concerning the experimental data, it was not possible to find a trustable gait database. Initially it was intended to use data from a public database so a comparison of results could be made, but it was not possible since such database was not found. Biomechanical Laboratories were contacted but any of them showed availability to share data. Finally, it was not possible to validate *ExoBody* model because it was not possible to measure the ground reaction forces.

With this in mind, there are several things that can be done to continue this project.

First, the use of a specific CAD modelling of the H1 exoskeleton would make the model more accurate which would lead to in more truthful results in the simulations. Consequently, it would improve the reliability of the *Exoskeleton* model.

The *ExoBody* model must be validated. Analog to the validation of the *Exoskeleton* model, static and motion trials should be performed in order to collect experimental data, though in this case, the trials have to be done with several healthy people. Furthermore, the validation of this model requires the verification of ground reaction forces. After having the ExoBody model validated, it will be a powerful tool that will allow researchers and clinicians to benefit from it and improve the treatment design and clinical decision.

One of the limitations of this work was the impossibility to find a public data base about gait disorders. Thus, starting from the Gait Pattern characterization presented in this work, would be very challenging to preform gait analysis of a large number of individuals, healthy controls and with gait disorders, and release all this work in a public data base.

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